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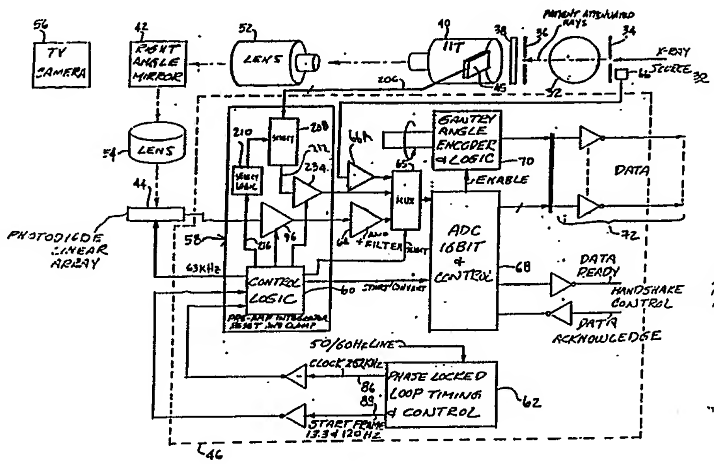
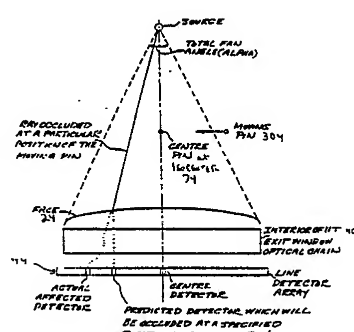
(57) Abstract





A computed tomography method and apparatus for sequentially projecting partial fan beams through a sample (32), measuring data resulting from each projection, and generating an image of a sample section from the measured data without reordering the measured data to simulate data obtained by projecting parallel ray beams through the sample (32). Partial fan beam radiation received at a detector array (44) is processed to image a larger diameter sample section than could be imaged with full fan beams projected from the same source to the same detector array (44) with the same source-to-detector spacing(s). Preferably, X-ray radiation is transmitted through the sample (32) and the system includes an image intensifier tube (40) for receiving data representing at least a portion of the transmitted radiation. Preferably, the invention includes a linear detector array (44) which receives the image intensifier tube output and an extension detector array (45) which receives partial fan beam radiation which has propagated through the sample but which is not incident at the image intensifier tube. The data measured during each partial fan beam scan are preferably weighted to simulate full fan beam data. The weighted partial fan data signals are then processed in the same way as if they were unweighted data signals resulting from exposure of the sample (32) to full fan beams.

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PARTIAL FAN BEAM TOMOGRAPHIC APPARATUS AND DATA RECONSTRUCTION METHOD

Cross-reference to Related Application

U.S. Patent Application Serial No. ____________, filed on the same day and having the same assignee as the present application, discloses an apparatus in which the invention of the present application may be embodied, and which may used to perform the method of the present application. U.S. Patent Application Serial No. ________________ is incorporated herein by reference.

Field of the Invention

This invention pertains to a method and apparatus for obtaining a computed tomography image of a sample, by scanning the sample with partial fan beams.

Background of the Invention

Within recent years, much interest has been evidenced in a field now widely known as computed tomography. In a typical procedure utilizing computed tomography (or CT), an X-ray source and detector are physically coupled together on opposite sides of the sample portion to be examined. An X-ray beam is made to transit across the sample to be examined, while the detector measures the beam transmitted at a plurality of transmission paths defined during the transit process. The paired source and detector are then rotated into a different angular orientation with respect to the sample, and the data collection process is repeated.

Typically, a very large number of attenuation values are measured by procedures of this type, and the relatively massive amounts of data thus

accumulated are processed by a computer, which typically does a mathematical data reduction to obtain attenuation values for a very high number of transmission values (typically in the hundreds of thousands) within the section of the object being scanned (typically, a human body). This data may then be combined to enable reconstruction of a matrix (visual or otherwise) which constitutes an accurate depiction of the density function of the sample section examined. Skilled diagnosticians, in conjunction with medical applications, by considering one or more of such sections, may diagnose various body elements such as tumors, blood clots, cysts, hemorrhages and various abnormalities, which heretofore were detectable, if at all, only by much more cumbersome and, in many instances, more hazardous (from the viewpoint of the patient) techniques.

While first generation CT systems are powerful diagnostic tools, and were deemed great advances in the radiography art, such systems suffer from many shortcomings. This type of CT system often requires an undesirably long period of time for raw data acquisition, which among other things subjects a patient to both inconvenience and stress. The patient's inability to remain rigid for such a lengthy period, also could lead to blurring of the image sought to be obtained.

U.S. Patent 4,149,247, U.S. Patent 4,149,248, and U.S. Patent 4,149,249, each issued April 10, 1979 and assigned to the assignee of the present application, disclose apparatus and methodology which alleviate a number of the problems with first generation CT systems, most notably including the lengthy period previously required for computer

processing of the raw data provided by the detectors. The apparatus disclosed in the referenced patents utilizes a fan beam source of radiation and performs convolution method of data reduction, with no intervening reordering of fan rays, to eliminate errors and delays in computation which would otherwise result from such reordering.

The radiation source and detectors are positioned on opposite sides of the sample to be imaged, and are made to rotate through a complete or partial revolution around the sample while the detectors measure radiation transmitted through a section of the sample at the plurality of transmission paths defined during the rotational process.

In U.S. Patents 4,149,247, 4,149,248, and 4,149,249, the entire sample section to be imaged is exposed to the fan beam (in the same way that sample section SS in Figure C, to be described in greater detail below, is exposed to fan beam F). A fan beam of sufficient width to expose the entire sample section is denoted herein as a "full" fan beam. A fan beam having width insufficient to expose the entire sample section is denoted herein as a "partial" fan beam.

It would be desirable to design a CT system in which the sample is sequentially exposed to partial fan beams (rather than full fan beams) each having a different beam orientation, and an image of a sample section is generated from the resulting measured "partial fan" data. Radiation from each such partial fan beam, received at a detector positioned a first distance from the beam source, can be used to image a larger diameter sample section than could be imaged with a full fan beam using the same source-detector

pair separated by said first distance. However, until the present invention, it had not been known how efficiently to process "partial fan" data to generate a sample image having the same quality achievable by processing "full fan" data.

X-ray CT scanners have become a common tool for the diagnostic radiologist. Typically these devices are expensive (they are often sold at prices in excess of one million dollars), and have scan times of 1 to 2 seconds with 0.3 mm spatial resolution. Density resolution as low as 0.25% with degraded spatial resolution is achievable. The technology of generator/detector design and the improvements in the microcomputer area over the past 10 years have enabled image detection and processing to approach real time.

Radiation therapists often attempt to use scans from diagnostic CT scanners in planning a treatment, however, the relative position of organs within the body are not the same with respect to the CT X-ray source as when a patient is placed on a flat couch of the radiation therapy machine. This occurs because the diagnostic CT scanner couch is more crescent shaped. Therefore, radiation therapy simulators have come into use with patient couches that are identical to treatment couches. Also, in a simulator, the X-ray focal spot for fluoroscopic/radiographic imaging is positioned to allow the same target to patient isocenter as in the therapy machine. Beam shaping devices and other accessories can be added which attempt to duplicate the therapy setup exactly. Thus, the simulator yields a projected planar image of the patient anatomy that is much more geometrically compatible with the position of the therapy accelerator system. In addition to the

properly oriented radiographic information, if cross-sectional CT images could be obtained at the same time, then the therapist would be further aided in planning the treatment.

A radiation therapy simulator is a diagnostic imaging machine shaped to simulate the geometry of radiotherapy treatment units. A simulator includes an imaging source for generating imaging radiation (typically, X-ray radiation), a gantry to support and position the imaging source, a couch to support the patient, and an image forming system. The dimensions of the gantry are such that it positions the imaging source relative to the couch in a geometry similar to the geometry of a radiotherapy machine. Images formed on the imaging system can then be interpreted in terms of the geometry of the radiotherapy machine. Images can be taken from different angles to aid in the planning of how to form the radiotherapy beam to maximize dose to the target and minimize damage to healthy organs.

Because the geometry of the simulator attempts to simulate that of the radiotherapy machine, the imaging source and image forming system are limited to a configuration which is less than optimal for the quality of the image. Both the source and the image detector part of the image forming system are far from the patient, so that use of a partial fan beam would be particularly advantageous in a simulator system.

In conventional simulators, the image at the detector has been recorded on film, an image intensifier has been used to increase the brightness of the image which can be used to produce a television image, and a computer has been used to process and enhance the television image.

In the prior art, it is known to form a computed tomography image based on data obtained from a TV camera using an image intensifier tube (IIT) between the patient and a television camera. The output signal from the television camera is processed to form a digital signal which is further processed in a computer to form a tomographic image. This prior art system employing the television camera produces a noisy image of marginal value in simulation and planning.

Similar CT attempts using X-ray image intensifiers with video cameras have been made in the past by various groups. However, from prior CT experience, we believed that the use of video camera signal based on data off the IIT was one of the major limiting features in these designs. Compared to the IIT, conventional video cameras have horizontal spatial resolution of 3-4 line pairs per mm over a 30 cm field, but their intensity output is both limited and nonlinear. Typically, tube video camera instantaneous signal dynamic range is limited to only two or three orders of magnitude. Conventional solid state video cameras have good linearity spatially and in intensity, but their signal dynamic range is also limited to about one thousand at room temperature and with averaging lines possibly four thousand to 1.

In order to maintain X-ray photon statistics on a 16" (40 cm) diameter body, a detector with a minimum signal to noise ratio (S/N) of at least 200 thousand to 1 is necessary. This assumes a typical surface dose of 2 rads/scan and no compensating bolus around the patient. It is also necessary that the IIT, lens optics and photo detector yield an X-ray to electron quantum efficiency of greater than unity.

Medical or industrial X-ray CT typically requires a detector system with millimeter spatial resolution and photon limited intensity resolution. The rate of photons emitted from an X-ray source is statistical and follows a Poisson distribution. Thus, any ideal measurement of photon intensity has a root-mean-square [rms] noise equal to the square root of the average number of photons detected. Therefore, the detector system must have a total quantum detection efficiency [QDE] greater than unity to maintain photon statistics. Also, since additional random noise adds in quadrature, the detector electronics must have a rms input noise level below the photon noise.

Throughout the specification, including in the claims, the term "detector array" will be used to denote not only an array of two or more detectors, but also a single detector (having one or more discrete detecting elements).

Summary of the Invention

The invention is a computed tomography method and apparatus for imaging a sample by sequentially exposing the sample to partial fan beams (rather than full fan beams) each having a different beam orientation, measuring (at a detector array) "partial fan" data resulting from each exposure, and generating an image of a sample section from the measured data. Radiation from each partial fan beam received at the detector array is processed to image a larger diameter sample section than could be imaged with full fan beams using the same source and detector array separated by the same source-to-detector spacing(s).

Raw detector array data is generated as a result of each projection of a partial fan beam through the sample. The raw data is corrected, and then weighted in order to reduce artifacts resulting from calibration errors in those detectors which receive partial fan beam radiation that has propagated through or near the scan isocenter. The weighted data signals are then processed in the same way as if they were raw data signals resulting from exposure of the sample to full fan beams, to generate an image of a sample section.

In one embodiment, the weighting signal includes a sinusoidal middle portion concatenated between a first end portion having a constant zero value and a second end portion having a constant value of two. In another embodiment, the middle portion consists of a first sinusoidal portion, a second sinusoidal portion, a portion between the two sinusoidal portions having a constant value of 1.

Preferably, when X-ray radiation is transmitted through the sample, the inventive system includes an image intensifier tube (IIT) for receiving data from at least a portion of the transmitted radiation. X-ray photons incident at the IIT are absorbed by a thin (0.3mm) cesium iodide (CsI) scintillator on the face of the IIT. The CsI crystals emit light photons which are converted to electrons by the photocathode of the IIT which are accelerated and focused onto a phosphor on the exit window of the IIT for conversion back into light photons.

In one preferred embodiment, the detector array of the invention includes a linear detector array which receives the IIT output, and also an extension detector array which receives partial fan beam

radiation which has propagated through the sample but which is not incident at the IIT.

Brief Description of the Drawings

FIG. M is block diagram of an embodiment of the inventive system.

FIG. N is a simplified prospective view of a radiation treatment simulator in conjunction with which the present invention can be used.

FIG. A is a block diagram of a detector system employed in an embodiment of the invention.

FIG. B is a diagram of simulator geometry for a full fan beam (head) CT.

FIG. C is a diagram of the reconstruction diameter of the head.

FIG. D is a diagram of simulator geometry for a partial fan beam (body) CT.

FIG. E is a diagram of the reconstruction diameter of the body.

FIG. H is the moving pin arrangement used in the center finding procedure.

FIG. I is a plot of a typical detector response obtained in the center finding procedure.

FIG. L is a cross-sectional view of an embodiment of the invention which includes a detector array having a separate detector extension.

FIG. O is a simplified side cross-sectional view of a portion of the detector assembly of a preferred embodiment of the inventive system.

FIG. P is a block diagram of a portion of the detector assembly of a preferred embodiment of the invention.

FIG. Q is a schematic diagram of preamplification circuitry for processing raw signals

produced by a detector array of a preferred embodiment of the invention.

FIG. R is a timing diagram representing a data acquisition steps of a preferred embodiment of the inventive method.

FIG. S is a simplified perspective view of an extension detector array employed in a preferred embodiment of the inventive system.

FIG. T is simplified perspective view of detector of the FIG. S array.

FIG. U is a graph representing a sensitivity profile of the FIG. T detector.

FIG. V is schematic diagram of circuitry for processing raw signals produced by an extension detector array of a preferred embodiment of the inventive system.

FIG. W is a flow diagram of the steps performed in a preferred embodiment of the inventive method.

FIG. AA is a diagram representing simulator geometry during a partial fan beam scanning operation.

FIG. AB is a diagram representing simulator geometry during a partial fan beam scanning operation.

FIG. AC is a diagram representing simulator geometry during a partial fan beam scanning operation.

FIG. AD is a table of data of the type obtained during calibration of an embodiment of the inventive system.

FIG. AE is a representation of actual and desired angular intervals associated with a detector array of a preferred embodiment of the invention.

FIG. AG is a graph of the weighting assigned to data from projections that are obtained by

overscanning an object in a preferred embodiment of the inventive method.

FIG. AH is a diagram explaining selection of an appropriate weighting function W(x) for use in weighting partial fan data in accordance with the invention.

FIG. AI is a graph of another suitable weighting function W(x).

FIG. AJ is a diagram representing simulator geometry during a full fan beam scanning operation.

Lexicon

ABS: Automatic Brightness System, ADC: Analog to Digital Converter, CsI: Cesium Iodide,
CT: Computed Tomography, DMA: Direct Memory Access,
FAD: Focus to Axis [Isocenter] Distance,
IIT: Image Intensifier Tube, Pb: Lead,
PSF: Point Spread Function, RMS: Root Mean Square,
and TDC: Top Dead Center

Glossary

The following is a glossary of elements and structural members as referenced and employed in the specification:

30 Radiation beam source
34 Pre-patient collimator and beam shaping filters
36 Post-patient collimator
38 Anti-scatter grid
40 Image intensifier tube
42 Right angle mirror
44 Photodiode linear array
52 First lens
54 Second lens
56 TV camera
58 Preamplifier integrator reset and clamp circuit

62 Phase locked loop timing and control circuit
64 Amplifier and filter
66 Normalization detector
68 Analog to digital connector and control circuit
70 Gantry angle encoder and logic

Description of the Preferred Embodiments

In the inventive system, an object of interest is scanned to provide a series of projections at different selected angles about a rotational axis of the object. Each projection is formed by passing a fan beam of radiation through a portion of the object at one of the selected angles. For each projection, the object-attenuated fan of radiation is received by a detector array.

In a preferred embodiment of the invention, the x-ray radiation is employed to image the object, and the detector array is suitable for detecting x-ray radiation. However, it is contemplated that other types of radiation (with suitable detector arrays) may be employed to image an object in accordance with the invention. Throughout the application, including in the claims, the phrases "fan beam of x-ray radiation" and "fan beam" (and variations thereon) are used to denote fan beams of any type of radiation, including x-ray radiation, other electromagnetic radiation, and non-electromagnetic radiation.

In a preferred embodiment, the detector array includes an image intensifier tube for converting radiation photons to visible light photons. The visible light photons are then detected using a photodiode linear array. Signals from the photodiode linear array are conditioned and then converted into digital form.

The digital information is then processed under computer control to correct the data for background noise, non-linearities in the image intensifier tube and photodiode array, and for other effects. The corrected data from each projection are used to reconstruct an image of a cross section of the scanned object.

Multiple scans can be performed to provide a three-dimensional view of the object of interest. Each image resulting from these scans can be displayed on a monitor, with variations in the image being displayed representing different absorption coefficients or absorption densities. Each reconstructed image can also be stored in digital form for later viewing.

The use of a full or a partial fan-beam of radiation depends upon the diameter of the object being scanned and the dimensions of the detector available. For example, where a 12" image intensifier tube ("IIT") is used, and a scan of a small object (such as the head of a human patient) is desired, we prefer to use a "full" fan beam (which exposes the entire object section to be imaged).

When imaging with a full fan beam, the center of the full fan beam, the isocenter of the object (head) being imaged, and the IIT are lined up with one another.

On the other hand, when a large object (i.e., the body of a patient) is to be scanned and a 12" image intensifier tube (IIT) is to be used, the diameter of the body is often too large to be fully exposed to a full fan beam and for the transmitted portion of the full fan beam to be received within the 12" width of the IIT. In this case, a partial fan beam should be used, with the IIT offset from the

axis defined by the fan beam source and the isocenter of the body being imaged.

Imaging Extension and Partial Fan Beam

In an embodiment of the inventive system, the detector array includes an imaging extension array (extension array 45 of Figure M) which enables the system to image an object of maximum diameter of 50cm (in comparison with a maximum object diameter of 40cm for the system without such extension array). The extension array also allows for an increase in patient to gantry clearance of from 60cm to 77cm, and allows for a full 48cm wide patient table to be used.

When a 12-inch IIT is used without the extension array, the maximum size of the objects which can be scanned is limited by the IIT diameter and its distance from the x-ray source.

The scanning geometry will be described with reference to Figures B, C, D, and E. Each of Figures B and D is taken transverse to the plane of the quasi-planar, flat beam (beam F in Fig. B and beam P in Fig. D), so that the axis of rotation of rotating arm 12 (shown in Figure is in the plane of Figures B and D. Each of Figures C and E is taken in the plane of the beam, so that the axis of rotation of the rotating arm 12 is orthogonal to the plane of Figures C and E. In Figs. B and D, the focal spot of X-ray source 30 is positioned approximately 100cm from isocenter 74 of the object to be scanned. In Figure B, IIT face 24 is positioned approximately 32cm from isocenter 74. In Figure D, IIT face 24 is positioned approximately 40cm from isocenter 74.

Positioned in the beam path are beam dimension adjustment jaws 76, followed by pre-patient collimator 34 at approximately 65 cm (in Fig. B) or

55-65 cm (in Fig. D) from the focal spot of source 30. X-ray normalization detector 66 is positioned on the x-ray source side of the pre-patient collimator 34. Post-patient collimator 36 is positioned just above IIT face 24.

The active area of a 12-inch image intensifier tube (and thus of IIT face 24 shown in Figures B-E and M) varies from unit to unit, but typically has been 28cm ±1cm across the face at 135cm from the x-ray source. As shown in Figure C, for a full-fan scan of object SS (which may be a patient's head) using full fan beam F originating at radiation beam source 30 (whose focal point is a distance of 100cm from isocenter 74 of object SS), geometry limits the maximum diameter of the object to be about 21cm. This covers approximately 95% of the U.S. male population according to Diffrient's Humanscale publication. (See N. Diffrient et al., Humanscale 11213 Manual, 1979, The MIT Press.) Most diagnostic CT scanners have a 25cm maximum head scan circle which allows for approximately 100% population coverage and less critical patient positioning.

In accordance with the invention, both head and body scans can be increased in size by using a partial fan beam. The simulator head scan circle can be increased to 25cm by using a partial fan beam rather than a full fan beam. In a partial fan beam mode (represented by Figure E) one edge of IIT face 24 is shifted a few centimeters off the axis between source radiation source 30 and the isocenter of object SS. With a partial fan beam P as shown in Figure E, a 360 degree scan is performed by rotating source 30 and IIT face 24 through 360 degrees about the isocenter. In this partial fan mode, a larger area may be imaged compared to the full fan mode.

Some loss in contrast and spatial resolution results because the entire object is not viewed in each projection.

The partial fan mode projection data is then reconstructed using a technique to be discussed in detail below.

When performing body scans in the partial fan mode, the IIT is shifted to the maximum extent (with respect to the axis between the beam source and the isocenter of the body). The largest body that may be scanned with the described arrangement is 40cm diameter. This covers approximately 95% of males in the U.S. across the chest, but less than 50% across the shoulders. The majority of diagnostic scanners have a 50cm maximum body scan which covers 97% of the same population.

Simulator

In one application, the present invention is used in conjunction with a radiation treatment simulator and planning system.

Radiation treatment simulator and planning systems ("simulators") simulate the geometry and movement of megavoltage radiation therapy equipment. The following are the basic items into which a simulator can be divided: floor mounted drive unit with rotating arm, X-ray head and crosswire assembly, detector including image intensifier, treatment table, relay frame, and control units. Many of the basic simulator elements suitable for use in with the present invention can be found in the Ximatron CR Radiotherapy Simulator System, manufactured by Varian, the assignee of the subject application.

With reference to FIG. N, drive unit 10 typically comprises a welded steel fabrication which

is bolted on a plinth, which is preferably cast in the floor, prior to the completion of the final floor finish. The drive structure houses a variable speed electric drive unit and a high precision slewing ring bearing on which is fitted the rotating arm 12. On the arm 12 are mounted the carriages 14 and 16 for the X-ray head assembly 18, and the detector assembly 20, which preferably includes an IIT. Attached to the front of the arm is a circular disk 22, the circumference of which carries a scale mark in degrees from 0.0 to 360.0 degrees. A screen wall (not shown) is supplied which carries the zero data mark for the scale together with a small sub-scale for ease of reading if the zero data is visually obstructed. The screen wall is built into a partition wall which seals off the drive unit and control gear from the room, thus presenting a clean finish.

Protruding through the top of the arm 12 is the X-ray head assembly 18 which is carried on a rigidly constructed steel fabrication. The X-ray system on the simulator has a generator having a typical output of 125 kVp and 300 mA (radiographic mode) or 125 kVp and 30 mA (fluoroscopic mode), in conjunction with a double focus (0.6 mm and 1 mm) X-ray tube, with a permanent 2 mm element and filter. The X-ray tube is mounted in a yoke on the end of the steel fabrication.

Mounted below the tube is a lead-bladed collimator which can be manually set to give field sizes from 0 to 35 by 35 cm and 100 cm F.S.A.D. The collimator also contains a lamp operated by a switch on the side of the housing, which defines the area of the X-ray beam passing through the blades onto the patient's skin.

Mounted with and in front of the collimator is a crosswire assembly. This is fitted with two pairs of motorized tungsten wires to give any square or rectangular field from 4 x 4 cm to 30 x 30 cm at 100 cm F.S.A.D. Windows inside of the crosswire housing have scales fitted indicating the field sizes at 100 cm. These are repeated on electrical indicators fitted in the remote control console. The collimator and crosswire assembly have motorized and manual rotation over the range ±45°. A suitable scale is provided for reading the angular position. The complete head is capable of being electrically driven from its maximum F.S.A.D. of 100 cm down to 60 cm.

Protruding through the front bottom of the arm 12 is the detector assembly 20. This unit is mounted on a double carriage to enable scanning over an area of ±18 cm about the center of the X-ray beam both longitudinally and laterally. The complete assembly is also capable of being electrically driven from a maximum of 50 cm from the rotation axis to the IIT face 24, down to 10 cm. Anti-collision bars fitted to the IIT face 24 will, when operated, isolate the electrical supplies to the operating motors.

Table assembly 26 includes a steel framework supported on a large precision bearing ring. These are mounted in a pit cast in the floor. The framework carries telescopic ram assembly 28 for table 26 together with a circular floor section. The bearing allows table 26 to be isocentered ±100° about the X-ray beam, either electrically or manually. A scale is fitted around the edge of the pit for positioning.

Telescopic ram assembly 28 provides vertical movement of the table top from a minimum height of 60 cm to a maximum of 120 cm.

Attached to the top of the telescopic ram is a sub-chassis which provides for manual lateral movement.

Data Gathering and Signal Processing Elements

With reference to FIGS. A and M, a computerized tomography system according to the invention will next be described. FIG. A is a block diagram which illustrates the data gathering elements of a preferred embodiment of the inventive system in relation to an X-ray source 30 and patient 32.

FIG. M is a block diagram which illustrates elements (indicated by a "*") which are added to the above-referenced Ximatron simulator system to obtain a preferred embodiment of a CT simulator system in accordance with the present invention. In FIG. M, it can be seen that the additional elements include: pre-patient collimator 34, post-patient collimator 36, grid 38, IIT 40 (having face 24), right angle flip mirror 42, photodiode linear array 44, extension detector array 45, 16-bit ADC and interface electronics 46, data acquisition interface 48, and processing and display computer 50.

With reference now to FIG. A, x-ray source 30 passes radiation through pre-patient collimator 34, then through the patient 32, then through post-patient collimator 36 and anti-scatter grid 38 to image intensifier tube 40 (IIT) and imaging extension 45.

The image from IIT 40 is projected onto photodiode linear array 44 using first lens 52, mirror 42 and second lens 54. When mirror 42 is swung out of the way, the image from IIT 40 can be viewed with television camera 56.

Signals from photodiode array 44 and extension array 45 are sent, on command, to a pre-amp, integrator, reset and clamp circuit 58. The signals from array 44 are amplified in input stage 96 of circuit 58, and supplied to an input terminal of multiplexer 65.

Extension array 45 is connected to pre-amp, integrator, reset and clamp circuit 58 by way of a multiwire harness 206. Signals from each of the photodiodes in extension array 45 are brought out from, and a 5V reference and ground connections are provided to, extension array 45 over the multiwire harness 205. As was the case with the photodiodes of photodiode linear array 44, a low noise reference is used at the anode of each photodiode of extension array 45. The signals from the photodiodes of extension array 45 are then applied in parallel to a bank of select circuits 208. These select circuits 208 are controlled by select logic circuit 210 to sequentially and serially place the signals from the photodiodes onto a video line 212. The signals on line 212 are amplified in amplifier 234, and are supplied to an input terminal of multiplexer 65.

Control logic circuitry 60 in the preamp, integrator, reset and clamp circuit 58 provides timing signals, which are in turn derived from clocks provided by a phase locked loop timing and control circuit 62. Phase locked loop timing and control ircuit 62 is synchronized to the 50/60 Hz line frequency. Control signals and light intensity signals are sent from the pre-amp, integrator, reset and clamp circuit 58 through an amplifier and filter 64, and multiplexed in circuit 65 with amplified signals from x-ray normalization detector 66 and from extension detector array 45. The output of

multiplexer 65 is supplied to analog-to-digital converter and control circuit 68 (ADC). ADC 68 sends an enable signal to a gantry angle encoder and logic circuit 70, and both circuits 68 and 70 send signals to data acquisition interface 48 (shown in FIG. M), via an optically isolated data path 72. From interface 48, signals are sent to data processing computer 50 (shown in FIG. M). Computer 50 returns handshake signals through interface 48 to ADC 68.

Individual components of the described system are described in greater detail in referenced U.S. Patent Application Serial No. __________, which is incorporated herein by reference.

For a "head" scan, a full fan-beam is used with a beam thickness of approximately 5mm (at isocenter 74). IIT face 24 is centered in the beam. Given the separation of X-ray source 30, isocenter 74, and IIT face 24 described above with reference to Fig. B, pre-patient collimator 34 has a slit width of approximately 5mm. Post-patient collimator 36 has a slit width of approximately 8mm. Beam dimension adjustment jaws 76 are set to provide a beam thickness at pre-patient collimator 34 wide enough to illuminate x-ray normalization detector 66, and to provide a beam width which is approximately 21.1cm at the isocenter 74 (see FIG. C). Further, beam dimension adjustment jaws 76 are positioned close enough to pre-patient collimator 34 so that the latter is the primary collimator of the beam.

For a large object ("body") scan, a partial fan beam is used, with a beam width of approximately 1 cm at isocenter 74. As can be seen in FIG. D, pre-patient collimator 34 is positioned between 55-65cm from the focal spot of X-ray source 30, and has a slit width of approximately 6mm. IIT face 24 is

positioned approximately 35cm from the isocenter 74. Post-patient collimator 36 has a slit width of approximately 13mm. As can be seen in FIG. E, with the axis of rotation of the rotating arm 12 is coming out of the plane of the paper, IIT face 24 is offset from center, and beam dimension adjustment jaws 76 are set so that a partial fan-beam is generated. For example, the beam from pre-patient collimator 34 would illuminate IIT face 24 edge-to edge, but the portion of the beam passing through isocenter 74 would be incident approximately 3cm from one edge of IIT face 24, as shown in FIG. E.

In angular terms, the beam would have an outer edge approximately 1.27 degrees from a center line 75 running between the focal spot of the X-ray source 30 and isocenter 74, and its other outer edge at approximately 10.49 degrees from the center line 75.

In the current embodiment, the slice thickness for head scans is 5mm (at isocenter) and the focus to intensifier distance (F.I.D.) is 145 cm. For body scans, the slice thickness is 1 cm (at isocenter), and the F.I.D. is 147 cm (which provides additional patient circle scan clearance).

A 14:1 cylindrically focused grid is preferably included in post-patient collimator 36. Pre-patient collimator 34 gives a well defined fan which helps to reduce patient dose and scatter. It also carries the beam shaping filters which attenuate the peripheral portions of the x-ray beam which pass through thinner portions of the patient 32. This not only reduces patient dose but also reduces the dynamic range over which IIT 40 and photodiode linear array 44 must respond.

FIG. L illustrates preferred dimensional relationships when the system's detector assembly

includes imaging extension array 45 as well as IIT 40. In such a configuration, a partial fan beam can used for a 50cm diameter patient scan circle. Pre-patient collimator 34 is positioned between 59-63 cm from X-ray source 30; and the isocenter is approximately 100 cm from X-ray source 30. IIT face 24 is positioned approximately 147 cm from X-ray source 30. The most vertical point of imaging extension 45 is about 8.5cm above IIT face 24.

With reference again to Figure D, X-ray normalization detector 66 mounted to one side of pre-patient collimator slit 34 and provides readings of source intensity which are used to normalize x-ray tube output variations during the scan. X-ray normalization detector 66 measures the unattenuated x-ray beam flux, allowing the detector to sample quite a large solid angle of the beam.

IIT 40 (which may be a conventional 12-inch medical image intensifier tube) serves as an X-ray photon to visible light photon converter. FIG. O is a cross sectional view of a typical embodiment of IIT 40, and the associated optics of the present invention. Incident X-ray photons are absorbed by the thin 0.3 mm CsI scintillator 78 behind IIT face 24. The CsI crystal emits light photons which are converted to electrons by attached photocathode 80. The electrons are accelerated and focused by focusing grids G1, G2 and G3, onto the output phosphor 82 for light conversion. The quantum efficiency of this process is characterized by a 4 to 5 order of magnitude increase in light photons to incident X-ray photons.

The CsI scintillator 78 is typically 12 mils thick. The output phosphor 82 is preferably "P20"

type (ZnCdS). The accelerating voltage is typically between 30-35kV. A one inch diameter image is produced at the output of IIT 40. Lead post-patient collimator 36 (FIG. A) and anti-scatter grid 38 (FIG. A) are used to define the CT slice thickness and to reduce X-ray scatter. Post-patient collimator 38 arrangement is mounted on a circular aluminum plate which is then bolted onto the mounting ring of IIT 40.

The overall QDE of the measurement system is dependent on the efficient collection of the light photons at the IIT output phosphor 82. The light collection optics 84 which views the IIT output is a lens system as shown in FIGS. A and O. The light collection efficiency for this lens geometry is proportional to the transmission and the square of the numerical aperture. With both lenses 52 and 54 focused at infinity, their light collection efficiency is approximately 1%, depending on the f-stop setting of the second lens. The IIT to light detector QDE in this case is still 2 to 3 orders of magnitude above unity.

In order to allow both fluoro TV camera 56 and photodiode linear array 44 to be permanently mounted, a dual port distributor with a motorized 45 degree flip mirror 42 is mounted on IIT 40 in place of a standard distributor. The flip mirror 44 normally rests in the fluoro position and when the CT mode is selected, the mirror is flipped through 90 degrees so that the IIT light output is directed onto linear detector array 44 through lens 54.

A variety of solid-state arrays have been evaluated for their suitability as detector array 44 for receiving image data from IIT 40. A commercially available 512 channel linear silicon diode array has

yielded excellent results when used as array 44. This array is linear image sensor number S2301, manufactured by Hamamatsu of Hamamatsu City, Japan. The array is 1 inch (25.6 mm) in length and 2.5 mm in width. Each diode detector is 50 microns by 2.5 mm with 72% active area. With the IIT output image at one inch in diameter, a 1-to-1 arrangement light collection optics 84 is used between IIT 40 and the array 44. Photodiode linear detector array 44 is built into a camera housing which is mounted at one of the exit windows of the right angled flip mirror 42.

The normalized photon response of the array is greater than 60% from 475 to 875 nanometers and overlaps the IIT output phosphor spectrum. The IIT light output spectrum from the "P20" phosphor peaks at 532 nm. Thus, the silicon photodiode spectral response is a reasonable match to the "P20" phosphor curve. The QDE for silicon is approximately 0.6 - 0.7 electrons/photon. Therefore, from the above, the overall QDE ratio of incident X-ray photon to electron-hole pairs in silicon is still much greater than unity for the system as a whole.

To obtain a spatial resolution of 1 mm on an object, the detector must have a sufficient number of channels to permit one to digitize the image. The photodiode linear array 44 has 512 channels, which translates to 0.6 mm per detector at the IIT face 24 for a 12 inch tube. Tests and specifications on IIT 40 have indicated that its spatial resolution is approximately 3.5 line pairs/mm over the 12-inch diameter, which exceeds the diode array's equivalent 0.9 line pairs/mm. Therefore, projected to the object, 1 mm resolution is possible in the reconstructed image data.

Each channel of photodiode linear array 44 is capable of accumulating 22 pico coulombs of charge during an exposure. The noise characteristics of the commercial array and preamp is specified as 3500 electrons rms. The saturation level to noise ratio yields a single measurement maximum signal to noise ratio of 39,000:1.

The point spread response of the IIT 40 indicates a dynamic signal range of at least 100,000:1. The single channel dynamic range of the photodiode linear array 44 has been measured to be 35,000:1 when used with the manufacturer's preamp. Thus, using a single measurement, the photodiode linear array 44 does not have sufficient dynamic range to match the IIT 40 output.

During data gathering, we prefer to employ a dual exposure time scheme, and some additional improvements in the charge preamplifier circuitry, to permit the 100,000:1 signal range of IIT 40 to be utilized. A new preamplifier integrator design provides a 50,000:1 range at room temperature. The combination of the dual exposure time scheme and new preamplifier integrator design provides an array-preamp dynamic range of 400,000:1, or 19 bits total, for each channel, over a measurement interval of 100/83 msec (50/60 Hz) while maintaining photon statistics.

The improvements in dynamic range are obtained in part by minimizing of the effect of the charge amplifier reset noise in the preamplifier, by phase locking the measurement to the line frequency, and by using an analog amplification scheme to amplify the signal from photodiode linear array 44 prior to converting it to digital form.

One source of degradation in the dynamic range of the measurement electronics is line frequency related ripple and harmonics in the x-ray source. The ripple and harmonics are a by-product of the rectification of the line voltage used in generating the high voltage CW for the x-ray source.

As can be seen in FIG. A, phase locked loop timing and control circuit 62 provides a number of clocks which are synchronized to the line frequency. More specifically, phase locked loop timing and control circuit 62 includes a voltage controlled oscillator (not shown) which is operating at a preselected multiple of the line frequency and which is synchronized to the line frequency. Within phase locked loop timing and control circuit 62 are divider circuits which divide down the voltage controlled oscillator signal into a sampling clock 86 and a start frame clock 88. In the embodiment shown in FIG. A, the sampling clock 86 is 262KHz and the start frame clock includes a 13.3Hz component and a 120Hz component. These clocks are applied to pre-amp integrator, reset and clamp circuit 58. As will be described hereinbelow, these clocks are used in the sampling of the photodiode linear array 44 and the dual exposure time scheme. Furthermore, phase locked timing and control circuit 62 supplies a select signal to analog-to-digital converter and control circuit 68 to synchronize its operation. When the timing in the measurement electronics is phase-locked in the above manner, substantial rejection of the line frequency ripple and harmonics can be obtained.

Referring now to FIGS. P and Q, the pre-amp, integrator, reset and clamp circuit 58 will be described in greater detail. FIG. P includes a simplified schematic of photodiode linear array 44.

The anodes of the 512 photodiodes are connected to a low noise reference, such as a bandgap voltage reference. The cathode of each of the 512 diodes is coupled to a video line 90 by way of pass transistors 92. The pass transistors are sequentially pulsed by an internal clock which operates off of sample clock 86, FIG. A.

When a pass transistor 92 is pulsed, the charge which has accumulated on its associated photodiode is placed on video line 90. This charge is transferred onto capacitor 94 which is in the feedback loop of the input stage 96 of pre-amp, integrator, reset and clamp circuit 58. Input stage 96 operates as a charge amplifier, and provides at its output a voltage proportional to the amount of charge present on capacitor 94. Capacitor 98 couples the voltage at the output of input stage 96 to low pass filter 100, which provides a gain of approximately four. Low pass filter 100 has a high impedance input and acts as a prefilter prior to analog to digital conversion by ADC 68. ADC 68 is a single 16-bit linear analog-to-digital converter.

The pre-amp, integrator, reset and clamp circuit 58 includes a reset transistor 102 connected in parallel with capacitor 94, and a clamp transistor 104 connected to the end of coupling capacitor 98 connected to low pass filter 100. Reset transistor 102 is pulsed to discharge capacitor 94 in preparation for receipt of charge from the next photodiode being sampled.

It has been found that a random offset voltage is coupled into the signal path by way of capacitive feedthrough from the gate of reset transistor 102. This offset can be on the order of one-half the control voltage being applied to the gate of reset

transistor 102. It has also been found that the addition of clamp transistor 104 reduces the above offset by a factor of five.

In operation, reset transistor 102 is pulsed with a positive going pulse for a predetermined time, such as two microseconds. At the same time, clamp transistor 104 is pulsed with a negative going pulse, but for a period about twice as long, such as four microseconds. During the time the clamp transistor 104 is pulsed, coupling capacitor 98 charges to the offset voltage. When the negative going pulse is completed, the end of coupling capacitor 98 connected to low pass filter 100 follows the output of input stage 96, which will assume a voltage proportional to the charge being transferred from the next photodiode being sampled in photodiode linear array 44.

FIG. Q provides a detailed schematic of the circuitry used to implement one embodiment of the pre-amp, integrator, reset and clamp circuit 58. In this embodiment, low pass filter 100 is implemented in three separate stages: 106, 108, and 110, with coupling capacitor 98 and clamping transistor 104 being located between stages 106 and 108. Stage 106 is non-inverting and provides a gain of 3.6; stage 108 operates as a follower; and stage 110 in inverting and provides a gain of 1.2.

Also shown in FIG. Q are circuits 112 for generating a three-phase clock for used in sampling photodiode linear array 44; the reset pulse 114 and clamp pulse 116 supplied to reset and clamp transistors, 102 and 104, respectively; and also the convert signal 118 supplied to ADC 68.

Referring to FIGS. S, T and U, imaging extension detector 45 includes an array of 32 discrete detectors 200. Each detector includes a high density

cadmium tungstate ($CdWO_4$) scintillating crystal 202 mounted and optically coupled to a UV-enhanced silicon photodiode 204. The scintillating crystals 202 are 2mm wide by 12mm long by 3mm deep, and have the following characteristics:

Stopping power of 150keV gammas in 3mm..90%
Stopping power of 3MeV gammas in 12mm...30%
Light output relative to NaI (T1).......40%
Wavelength of maximum emission..........540nm
Decay constant.........................5μsec
Afterglow at 3msec.....................0.1%
Index of refraction at 540nm............2.2-2.3
Temp. coeff. of light out at 300K.......0%/deg.K
Density................................7.9g/cc
Melting point..........................1598 K
Hygroscopic............................No

These crystals are available from NKK of Tokyo, Japan, and from Harshaw Chemical of Ohio.

Each crystal is resin mounted to a photodiode 204, with the side facing the photodiode being polished, painted with a white reflective coating, and sealed with black epoxy. The photodiodes 204 are preferably model no. S1337-16Br manufactured by Hamamatsu of Japan, and have the following characteristics at 25 degrees C:

Quantum efficiency at 540nm.....70%
Radiant sensitivity at 540nm...0.35A/W
Noise equivalent power.........$6x10^{-15}$W/root Hz
Rise time......................0.2μsec
Dynamic range..................10^{-12} to 10^{-4} A
Dark current at 10mV rev. bias.25 pA max.
Junc. cap. at 10mV rev. bias...65pF
Dark current at 5V rev. bias...60 pA typ.
Junc. cap. at 5V rev. bias.....22pF

Photodiodes 204 have an active area of 1.1 by 5.9 mm. The photodiode 204 case is 2.7mm wide by 15mm long. This yields detector spacing of 2.85 and the 9mm length of the imaging extension detector array 45. Each scintillating crystal 202 has an active face of 2mm by 12mm, which yields a slice thickness of the imaging extension detector array 45, referred to the isocenter, of 8mm.

The limitation on photodetector length is due to the photodiode's active length and case size. The x-ray signal is typically very high at the periphery of the scan circle and thus the loss in signal is insignificant. More specifically, the dynamic range requirements at the periphery of the scan circle are a factor of 10 less as compared to detectors at the center. Detectors at the center typically receive the fewest x-ray photons. Further, the width of the imaging extension detector 45, referred to the isocenter, is approximately 1.9mm, compared to 0.37 mm for the photodiode linear array 44. This yields a spatial resolution approximately a factor of 5 less than the photodiode linear array 44. However, this lower spatial resolution is not significant because high spatial content objects are typically not viewed at the periphery of a body scan circle.

Extension array 45 is mounted at the edge of existing 30cm IIT face 24. A measured x-ray sensitivity profile along the face of one of the detectors of array 45 is shown in FIG. U. The combination of IIT 40 and imaging extension detectors 45 forms an overlapping hybrid detector design.

X-ray photons, which have passed through the object being scanned, are incident on the 2mm by 12mm face of the scintillating crystals 202. Each photodiode 204 is operated in the photoconductive

mode with 5 volts of reverse bias applied after each integration and readout. X-rays absorbed in the scintillator produce light photons which are converted to electron-hole pairs in the diode. The resulting current flow discharges the diode and the preamplifier measures this loss in charge for each channel. This scheme is a discrete circuit implementation that is similar to the approach used in large integrated circuit linear and 2-D photodiode arrays.

A multichannel scanning charge preamplifier is used to sample each of the 32 detectors and multiplex this data with the existing IIT camera 512 channel data. The sampling of these detectors operate similarly to that used in connection with photodiode linear array 44, described hereinabove. Referring to FIGS. A and V, a simplified block diagram and a detailed schematic of such a preamplifier circuit are provided.

Imaging extension detector array 45 is connected to pre-amp, integrator, reset and clamp circuit 58 by way of a multiwire harness 206. Signals from each of the photodiodes 204 in the imaging extension detector array 45 are brought out from, and a 5V reference and ground connections are provided to, imaging extension detector array 45 over the multiwire harness 206. As was the case with the photodiodes of the photodiode linear array 44, a low noise reference is used at the anode of photodiodes 204. The signals from the photodiodes 204 are then applied in parallel to a bank of select circuits 208. These select circuits 208 are controlled by select logic circuit 210 to sequentially and serially place the signals from the photodiodes 204 onto a video line 212.

Select circuits 208 are gating circuits which transfer data applied at their "D" inputs to the "S" output, under control of signals applied to the "G" terminals. Select logic circuit 210 includes a bank of parallel to serial shift registers which operate to scan the bank of select circuits 208. Select logic circuit 210 can be implemented using part number 74HC164.

The scan is begun upon receipt of a start extension pulse on line 212. This pulse is clocked along the parallel to serial shift registers 210-A through 210-D, at a rate set by the extension clock supplied on line 214. As can be seen from FIG. A, the start extension pulse and extension clock are supplied from control logic 60 on line 216.

As was the case with the photodiode linear array 44, the preamplifier for extension detector array 45 uses an input amplifier 219 with a capacitor 220 connected in a negative feedback configuration. Line 212 is tied to one end of capacitor 220. A reset transistor 218 connected in parallel with a capacitor 220 to reset in preparation for receipt of the next sample. A coupling capacitor 224 couples the output of input amplifier 219 to non-inverting amplifier 226. A clamp transistor 222 is connected to the end of coupling capacitor 224 that is connected to non-inverting amplifier 226. Finally, the output of non-inverting amplifier 226 is connected to lowpass filter 228.

Reset transistor 218 is pulsed to discharge capacitor 224 in preparation for receipt of charge from the next photodiode being sampled, and clamping transistor 222 is pulsed during this operation, as is the case with reset transistor 102 and clamping transistor 104 in the charge amplifier for the

photodiode linear array 44. The reset pulse for reset transistor 218 is provided on line 230, while the clamping pulse is supplied on line 232 from the extension clock.

Referring now to FIG. R, the relative timing involved in the data acquisition will be described in greater detail. For a partial fan beam scan, a complete set of X-ray transmission data is obtained by collecting at least 720 projections (two per degree of rotation) during a period of 60 seconds. In practice, slightly more than 720 projections are collected and the gantry is rotated through slightly more than 360 degrees. However for purposes of the following discussion, 720 projections and 360 degrees of rotation will be assumed.

In FIG. R, line 120 illustrates the allocation of the 720 projections across one revolution of the X-ray head assembly 18 and IIT assembly 20 about the object of interest. Each such projection takes up approximately 83.3msec (60 Hz). Line 122 illustrates how each projection can be viewed as a series of periods of duration T. The example illustrated in FIG. R uses T equal to 8.33msec. Within a projection, the periods are grouped to define a long sample interval, 9T in length, and a short sample interval, 1T in length.

It has been found that the dynamic range of the photodiode linear array 44 can be greatly increased by utilizing a dual exposure time scheme. That is, by using a short sample interval and a long sample interval over which the photodiodes are permitted to convert photons to electrons, the most accurate measurement of the two can be selected for use.

As discussed briefly above, the point spread response of the IIT 40 indicates a dynamic signal

range of at least 100,000:1. One the other hand, the single channel dynamic range of the photodiode linear array 44 has been measured to be 35,000:1 when used with the manufacturer's preamp. As such, the individual photodiodes will saturate under high levels from IIT 40. With a two-interval sampling approach, the short interval sample will be the most accurate for high intensity levels from IIT 40, and the long interval sample will be the most accurate for low intensity levels from the IIT 40. This effectively extends the dynamic range of the photodiodes into the 100,000:1 range.

In practice, for the commercially available photodiode array identified hereinabove, the saturation level is 22 picocoulombs. The f-stop of second lens 54 is set so that the photodiodes will not saturate over a short interval period when there is no object in the x-ray beam; i.e. with the beam at 125kVp and 15mA. Preferably, this f-stop setting will result in a light level at the photodiode linear array 44 of about one-half to three-fourths the saturation level with no object in the x-ray beam.

Returning now to FIG. R, the timing for the long and short interval sampling will be discussed in greater detail. Line 124 illustrates the points at which the photodiodes of photodiode linear array 44 are sampled relative to one another. In the left most part of line 124 there is shown a first series 126 of 544 sample points: 512 for the photodiode linear array 44, and 32 for the imaging extension detector array 45. These occur during the left most T-period of line 122; i.e. during the last T-period of projection 1, line 120. In line 124, the second series 128 of 544 sample points occurs during the first T-period in projection 2, line 122. It is to

be noted that no sample points occur in line 124 until the ninth T-period of projection 2, line 122. Thereafter, a third series 130 of 544 sample points are shown occurring during the ninth T-period. Finally, a fourth series 132 of 544 sample points are shown occurring during the first T-period of projection 3.

The long interval sample for projection 2 is taken during the third series 130 of 544 sample points. The short interval sample for projection 2 is taken during the fourth series 132 of 544 sample points. For example, it can be seen from line 124 that the time period between the sampling of diode 1 in second series 128 and third series 130 is nine T-periods. Diode 1 is therefore permitted to integrate the incident photons for nine T-periods before it is again sampled. The samples taken in third series 130 thus represent the long interval sample for projection 2.

Conversely, the time period between the sampling of diode 1 in third series 130 and in fourth series 132 is only one T-period long. Thus, samples taken in the fourth series 132 represent the short interval sample for projection 2.

Line 134 in FIG. R illustrates the time for the sample points for diodes 1-4 in series 126, line 124. As can be seen, the time between sample points is approximately 15.3 microseconds. Within this 15.3 microsecond period, the input stage 96 of pre-amp, integrator, reset and clamp circuit 58 is reset (line 136), the clamp transistor 104 is pulsed (line 138), charge from the photodiode being sampled (for example photodiode 1) is placed on the video line 90 (line 140), and convert signal 114 is sent to ADC 68 (line

142). Lines 144 and 146 show the relative timing of the sampling pulses for photodiodes 2 and 3.

It is to be noted that there are four base-clock periods in each diode sample period. These base-clock periods are related to the 262KHz clock 86 from phase locked loop timing and control circuit 62, FIG. A. Similarly, the 8.33 microsecond duration of each T-period is related to the 120Hz clock 88 from phase locked loop timing and control circuit 62. Finally, the 13.3Hz clock 88 from phase locked loop timing and control circuit 62 relates to the duration of nine T-periods.

As was discussed above, the intensity of x-ray source 30 is monitored by x-ray normalization detector 66 (see FIG. A). The signal from x-ray normalization detector 66 is amplified and filtered in amplifier and filter 66A and then applied to a multiplexer 65. Also applied to the inputs of multiplexer 65 is the signal from pre-amp, integrator, reset and clamp circuit 58, and the preamplified output of extension detector array 45. The output of multiplexer 65 is then applied to ADC 68. A select signal is supplied to select between the signal from x-ray normalization detector 66, the preamplified extension detector output, or the signal from pre-amp, integrator, reset and clamp circuit 58 for conversion by ADC 68.

A gantry angle encoder and logic circuit 70 is attached to a potentiometer which measures the angle of the rotating arm 12. This encoder nominally gives ten (10) pulses per degree of gantry rotation and hence allows the determination of projection angles to 0.1 degree (12-bit counter).

Data acquisition interface 48 (shown in Figure M) is an optically isolated interface which utilizes

a conventional photodiodes and receiver link. Use of an optical link greatly reduces electrical ground problems.

In addition to the digital data and handshake signals from ADC 68 and gantry angle encoder and logic circuit 70, an analog channel (not shown) is brought out from amplifier and filter 64 via data acquisition interface 48 for use in calibration and set up purposes.

With reference to Figure M, processing and display computer system 50 preferably includes a conventional personal computer 50a (preferably a 80286 based personal computer), a conventional 20MFlop array processor 50b (such as a Vortex AT Array Processor available from Sky), a monochrome monitor 50c, a touch-screen color monitor 50d, and an image display card 50e (such as an image display card available from Matrox of Canada). Computer 50a and array processor 50b are provided with suitable external or internal memory, such as a 250 MB WORM (write-once-read-many) optical disk drive, 3 MB and 4 MB RAM memories, and a 30 MB hard disk.

Prior to data reconstruction, the data is preferably corrected for certain known error sources. For example, processing and display computer 50 preferably corrects for detector system spatial and intensity non-linearities and offsets. To minimize the effects of the point spread response of IIT 40, the data is preprocessed by array processor 50b. That is, after background subtraction and normalization, the array data is convolved with an empirical filter which compensates for the non-ideal point spread response. After all the projection data is obtained, a 512 x 512 pixel image is then generated using a convolution and back-projection

technique. The resulting CT image has better than 1 mm spatial resolution and 1% density resolution on a 20 cm water calibration phantom.

Possible sources of error in acquired data lie in both the imaging chain and in the mechanical system. Imaging chain sources of error (in no particular priority order) are: 1) time varying x-ray flux from X-ray source 30; 2) photon scatter; 3) IIT 40 (non-linearity across face, s-curve distortion, EHT variation with current, center detector, edge effects, curved face, dark current); 4) photodiode linear array 44 (non-linear response, saturation, long versus short integration values, dark current); and 5) optics (internal reflections, distortion, mirror alignment).

Mechanical system sources of error are: 1) wandering isocenter 74; 2) mechanical flexing; 3) non-uniform rotational speed; 4) lack of stiffness in IIT structure; and 5) non-repeatability of machine positioning.

The x-ray flux from X-ray source 30 can vary with time (power frequency fluctuations, photon statistics, etc.). This is measured directly by means of x-ray normalization detector 66. The output from x-ray normalization detector 66 provides a current proportional to the number of incident photons. It is assumed that this device is perfectly linear and the readings from the other detectors are normalized to it, i.e., detector elements are scaled as if the x-ray flux was constant and at its peak value.

Scatter of x-ray photons during their passage through the body is difficult to correct for. Our preferred approach is to attempt to eliminate scatter problems by: accurately collimating the fan beam, and

using a 14:1 cylindrically focused scatter suppression grid in front of the IIT face, although this does cause loss of primary x-ray photons.

Errors and distortion from the IIT 40 can arise for the following reasons: 1) uneven distribution of absorbing material/scintillator (CsI) at the IIT face; 2) curvature of the glass IIT face and its increasing thickness away from the center; 3) observed spatial non-linearity across the face of the IIT due to electron focusing errors; 4) s-curve distortion which varies as the IIT 40 is reoriented in the earth's magnetic field; 5) dark current (i.e., noise); 6) dynamic range (max signal:noise); and 7) finite point spread function across the tube face due to internal optical light scatter at the input and output of the tube.

Distortions and errors in the optical path are generally due to internal reflections and lens imperfections and can be summarized by the system point spread function.

For the system, overall light intensity is not a limitation, but x-ray photons are. The second lens 54 f-stop is generally set to 5.6 so it could easily be opened up to 4.0 to allow twice as many light photons through. The IIT 40 has a QDE of 1,000-10,000, therefore there can be a loss of light photons before the QDE of the system reduces to unity. Potential sources of errors in the detector array are: 1) non-linear detector/amplifier response with respect to number of light photons detected; 2) dark current; 3) different response with changes in integration period (for a given light input); 4) detector saturation; and 5) non-repeatability of center detector position.

A calibration procedure has been implemented to quantify and correct for data acquisition errors. The IIT 40, optics, and photodiode linear array 44 chain are treated as a single unit for the purposes of calibration and data correction. Information obtained as a result of these calibration steps is used to correct the data collected during an actual scan.

The calibrations are performed in the following order: a) mirror alignment; b) dark current (background); c) center detector and detector array limits (fan angle limits); d) detector system spatial linearity; and e) system point spread function. Additionally, every detector array is calibrated for response non-linearities by using a calibrated light source.

With reference now to FIG. H, the center detector of array 44 is identified as follows: 1) collect background data; 2) set the x-ray flux so that no detectors saturate; 3) collect data with no object in the beam path; 4) set a pin or needle at isocenter 74; 5) ignore detectors close to the ends of photodiode linear array 44 where readings fall off rapidly; 6) run a scan of the center pin to collect a full set of readings. FIG. I shows a typical set of readings for any particular projection.

The readings are then processed as follows: a) background correct the data; b) select long/short integration values (the longs will always be selected since the x-ray flux is set so that no detectors saturate); c) calculate ln(air norm)-ln(data); and d) set non-useful detectors to zero.

The results of the calculations are 0 (zero) for most detectors, with a positive attenuation value at the detectors which "saw" the pin 302.

"Find center" software searches for peak attenuation values for each projection and calculates a corresponding interpolated detector number, i.e., the center detector for that projection.

All the center detector values are then averaged to compensate for the possibility that the pin was not placed exactly at the isocenter 74 and for the flexing of the mechanical structure. The result is the identity of the center detector for the system, which is then saved for future use.

When a ruler with attenuating markers is placed over the face of the IIT and irradiated, if there are spatial non-linearities in the detector system, then the markers will not be equally spaced when viewed at the exit window of the IIT. In other words, although the photodetectors in photodiode linear array 44 are uniformly spaced, various effects in the IIT 40 and the imaging path can cause a detector, other than the predicted detector, to be affected when an object is positioned in the fan beam. See FIG. H.

Among the factors which influence spatial non-linearity are the curvature of IIT face 24. As can be seen from FIG. H, face 24 has a convex shape with respect to the fan beam. This results in rays at the outside of the beam striking the face 24 at a larger relative displacement than rays near the center of the beam. Within the IIT 40, non-linearities in the focusing grids G1, G2 and G3, can cause the trajectory of emitted electrons to depart from the predicted path. Lens errors in first lens 52 and second lens 54, and mispositioning of the right angle flip mirror 42, are also a source of spatial non-linearities.

In order to determine such system spatial non-linearity a second pin 304 is placed at an offset

from the center pin and then moved slowly through the beam. See FIG. H. In practice, this effect is actually achieved by keeping pin 304 fixed and rotating the gantry.

Referring to Figs. AA through AC, this effect is illustrated in simplified form. In each of Figs AA through AB, top dead center (TDC) of the gantry rotation path is shown at the top of the figure. The circle illustrates the path of the gantry rotation. X-ray source 30 is shown in a different position in each of the figures. For these different positions of X-ray source 30 it can be seen that a different detector in photodiode linear array 44 is affected by the occluded ray 306.

The angle θ, indicating the angle between center line 75 and TDC, is measured using gantry angle encoder and logic circuit 70. Angle α is the angular position of the calibration needle 304 from top dead center. The angle δ, can then be calculated from θ, α, the distance from X-ray source 30 to isocenter 74, and the distance between isocenter 74 and calibration needle 304, using well known geometric techniques.

From this data the angle of the occluded ray 306, and the detector which responded to the occluded ray 306, can be tabulated. See FIG. AD, where such a table is shown with data selected to illustrate the concept. From the geometry of the system, the detector which should have been affected for any position of the "moving" pin 304 can be predicted. This information is also included in the tabulation of FIG. AD. The total fan angle is calculated by examining when the "moving" pin moves into the fan beam and then leaves it.

In the reconstruction method of the present invention (to be described below), data corresponding

to rays of equal angular displacement are assumed. From the tabulated data (i.e., the data of FIG. AD), and corresponding detector readings taken during an actual projection, an intensity reading can be determined for each of the specific angles desired. An averaging/interpolation technique (to be described below with reference to Fig. AE) is performed to generate such "regularized" data by "moving" the actual detector readings into correct uniformly displaced "expected detector" slots for further processing (including back projection).

Measurements from extension array 45 are preferably combined with those from photodiode linear array 44 to provide a total of 544 measurements. Thus, when extension array 45 is attached to IIT 40, the first photodiode of extension array 45 is positioned to overlap the last several photodiodes in photodiode linear array 44. The detector spacing of the photodetectors in extension array 45 is approximately five times that of the effective detector spacing in photodiode linear array 44. However, this lower spatial resolution is not significant because high spatial content objects are typically not viewed at the periphery of a body scan circle.

"Regularized" data values which correspond to desired uniform angular displacements can be obtained from the extension array 45 data in the same manner that they are obtained from the photodiode linear array 44 data. This is accomplished by interpolating the raw measured data from array 45, and assigning the interpolated values to the correct, uniformly displaced "expected detector" slots ready for further processing (including back projection).

In a preferred embodiment (for each projection), 32 regularized data points from extension array 45 are combined with 512 regularized data points from photodiode linear array 44 to provide 544 regularized data points. By performing the averaging and interpolation process described below with reference to Fig. AE, adjacent points from this combined, regularized data set are then averaged to obtain a reduced set of data points which are then assigned to X "expected detector" slots (corresponding to X detectors equally spaced over the entire full fan beam detection area), for further processing (including back projection). Even in embodiments in which extension array 45 is not used, the data points from array 44 are preferably regularized, reduced, and assigned to the X "expected detector" slots in order to increase the spatial resolution of the image ultimately produced. In one class of preferred embodiments, X= 300.

The overall sequence of data collection and correction is shown in FIG. W. After the system is initialized (in step 343), data is collected in step 345.

During data collection step 345, the readings obtained at every projection are: Projection number; Short sampling interval value; Short normalizing detector value; Long sampling interval value; and gantry angular position.

In accordance with the preferred embodiment of the present invention, processing of the actual detector readings is performed while a scan is in process. Because a scan typically takes about a minute to complete, a significant amount of processing of the data can be done during the scan.

Step 347 is the first of the data processing steps. During step 347, the background level is subtracted from each detector reading as the data is received from each projection during a scan, and the data is corrected to compensate for nonlinearity of the detectors with intensity.

Also during step 347, the data from a scan (typically comprising approximately 800 projections in a 60 Hz system, or 650 projections in a 50 Hz system) are combined so as to reduce the data to a reduced data set representing a lesser number of projections (typically one projection per each degree of rotation around the gantry path). Such data reduction is accomplished by weighting the projection data values for projection angles close to each of the desired projection angles. The data reduction speeds the further data processing and helps to average out any noise.

Next, step 390 is performed to correct for the effect of the system point spread function by deconvolving the detector data.

Next, step 400 is performed to normalize the data. This is done by employing a phantom normalization and line integral difference calculation, which includes the steps of computing the natural logarithms of the corrected intensities measured during an actual projection, subtracting the natural logarithms of the normalizing data therefrom to generate difference signals, and subtracting the natural logarithms of intensities measured using a phantom having known absorption characteristics from the difference signals.

Next, in step 401, we perform overlap correction in the following manner, to be explained with reference to Fig AG.

In a preferred embodiment, the reconstruction algorithm assumes that 360 equally spaced projections have been obtained during rotation of the beam source and detector array through a 360 degree revolution about the object's isocenter. A scan typically results in data representing approximately 800 projections (in a 60Hz system) or 650 projections (in a 50Hz system) since, in practice, we prefer to rotate the gantry an additional 5 to 10 degrees beyond the top dead center of the gantry rotation path ("TDC") after 360 degrees of rotation. This results in a slight overscan. Projections are taken during this overscan region, as well as during the 360 degree rotation scan. As explained above, during step 347 the approximately 800 (or 650) projections are reduced to a lesser set of projections (for example 370 projections). During step 401, this reduced set of projections are weighted, to blend data from the overscan region with data from projections taken at the beginning of the 360 degree scan, thus producing data representing a total of 360 projections. Figure AG shows one suitable weighting function for data from 370 degrees of projections. From Figure AG, it can be seen that the data taken in the early and late projections (at gantry angles near zero degrees and near 370 degrees) are lightly weighted, while the data taken during the middle of the scan (for example, at 180 degrees) are "weighted" by a factor of one.

Next, step 402 is performed in which adjustments are made to compensate for geometric (spatial) non-linearities. As described above with reference to FIGS. AA, AB, AC, and AD, rays in the partial fan beam which are separated by uniform angles do not produce responses at detectors in the photodiode

linear array 44 which are spaced a correspondingly uniform distance apart.

FIG. AE illustrates the averaging/interpolation technique employed in step 402 of FIG. W, which corrects for these spatial non-linearities. The upper section of axis 308 illustrates the desired uniform angular interval between measurements, for example, a measurement every Δ degrees, between $\pm$ 12 degrees. The bottom section of axis 308 illustrates the actual angular interval between the actual measurements. Note that due to the spatial non-linearities in the imaging system detector responses occur at other than the predicted angles.

As can be seen from portion 310 of FIG. AE, intensity values for a desired angular position are determined by selecting a subset of the detector measurements and interpolating those measurements. Thus, for example, the intensity value for the angular position three-Δ intervals from the -12° point might be determined by interpolating the measurements from detectors 1 and 2. Similarly, the intensity value for the angular position two-Δ intervals to the left of the 0° position might be determined by interpolating the measurements from detectors 250-253. In the above manner, responding detector readings can be averaged/interpolated together and then "moved" into the required detector slots for further processing (including back projection).

Step 401 may be performed either before step 402 (as shown in FIG. W) or after step 402.

The corrected data is then written to the reconstructor input file. During step 404, corrected partial fan data from the input file is weighted (in a manner to be explained with reference to Fig.AH).

The weighted, corrected partial fan data (or non-weighted, corrected full fan data from the input file) is processed to reconstruct an image representing a cross-section of the scanned object. The resulting image is then displayed (step 406 of FIG. W).

Reconstruction and Back Projection Methods

We prefer to program the processing means of the invention (including computer 50a and array processor 50b) to operate in either a "full fan beam" reconstruction mode or a "partial fan beam" reconstruction mode.

In either mode, the data to be processed is generated as a result of scanning an object by sequentially projecting fan beams through a section of the object. Each fan beam has an orientation axis extending from the beam source through the object section's isocenter (for example, axis 75 is the orientation axis of beam P in Figure E). A partial fan beam must include the isocenter, but nevertheless has an orientation defined by an orientation axis through the isocenter.

In the full fan beam mode, raw data is measured by a detector array positioned symmetrically with respect to the isocenter, which receives radiation from full fan beams which have propagated through the entire object section. In the partial fan beam mode, raw data is measured by a detector array positioned asymmetrically with respect to the isocenter, which receives radiation from partial fan beams (each of which has propagated through less than the entire object section). Preferably (as shown in Figure M), detector array positioning and translation unit 41 is coupled to the assembly comprising IIT 40 and

extension array 45, for translating such assembly tangentially (relative to the gantry rotational path) between such symmetrical and asymmetrical positions.

In the full fan beam mode, we prefer to reconstruct an image of the scanned object section from the corrected full fan scan data (generated during step 402) in accordance with the correlation and back projection algorithms disclosed in above-referenced U.S. Patent 4,149,249, which patent is incorporated herein by reference. The input data for such full fan data reconstruction preferably consists of 300 data points for each of 360 projections.

In the partial fan beam mode, we prefer to reconstruct an image of the scanned object section from a weighted version of the corrected partial fan scan data generated during step 402. Before the data is weighted (as part of step 404), it has been regularized and reduced (assigned to X "expected detector" slots which correspond to X equally spaced detectors) in the manner explained above. Linear detector array D in Figure AH represents such X equally spaced detectors. In Figure AH, the parameter X is equal to M, and L is the number of such equally spaced detectors required to receive a full fan beam that exposes the entire object section 32 being imaged. The parameter L thus represents the width of the narrowest full fan beam which would span sample section 32, M represents the width of the partial fan beam projected on sample section 32, and the ratio M/L represents the "degree" of the partial fan beam.

In order to weight the 300 regularized, reduced data values from a projection in accordance with the invention, each data value should be multiplied by a factor corresponding to a weighting curve W(x) of the type shown in Figure AH. Thus, for example, data

value d_i should be multiplied by weight factor W_i, and each other data value should be multiplied by a value representing a different point on curve W(x).

The range of weighting curve W(x) is from x = 0 to x = L. The parameter x = L corresponds to the position of the rightmost detector in array D (the detector farthest from beam orientation axis 75, and hence farthest from a "center" detector positioned along orientation axis 75). The parameter x = L/2 corresponds to the position of the "center" detector in array D which receives partial fan beam radiation that has passed through isocenter 74 of object 32. The parameter x = L-M corresponds to the position of the leftmost detector in array D.

Weighting curve W(x) is preferably a smooth, slowly varying function, and should satisfy the following conditions:

1. W(0) = 0;
2. W(L) = 2; and
3. W(x) = 2 - W(L -x).

In the preferred embodiment shown in Figure AH, W(x) is a continuous function having three portions: a first portion equal to zero in the range from x = 0 to x = L-M; a sinusoidal portion in the range from x = L-M to x = M; and a third portion equal to two in the range from x = M to x = L.

In the alternative preferred embodiment shown in Figure AI, W(x) is a continuous function having three portions: a first portion equal to zero in the range from x = 0 to x = L-M; a non-decreasing portion in the range from x = L-M to x = M; and a third portion equal to two in the range from x = M to x = L. Such non-decreasing portion has a sinusoidal portion in region I, a portion having constant value one in

region II, and a second sinusoidal portion in region III.

More generally, it is preferred that W(x) have a first end portion with constant value zero, a second end portion with constant value two, and a middle portion which increases in value from zero to two over its range. The range of such middle portion increases with increasing degree M/L of the partial fan beam. The middle portion is non-decreasing but need not be sinusoidal.

To appreciate the reason for weighting the partial fan data using function W(x), it should be recognized that during a full fan beam scan, data representing each beam path segment through the sample is measured twice: once during a first projection, and again during a second projection with the beam source and detector at a second angular position along the gantry path. Thus, in Figure AJ, if full fan beam source 30 projects a full fan beam through object section 32 having isocenter 74, the radiation propagating along path 77 is received at detector D_1 (of detector array A). Also, when the source and detector array A are rotated to positions 30' and A', respectively, the data measured at detector D_2 represents radiation propagating along the same path 77 through the sample. Thus, two data values are collected (at two different detectors) for each path 77 through the sample. Accordingly, the operation of weighting partial fan beam data obtained in accordance with the invention with weighting function W(x) compensates for the fact that a reduced amount of fan beam data is obtained with a partial fan beam scan (since partial fan beam data includes only one data value for each beam path through the

sample, unlike full fan beam data which includes two such values).

We have recognized that the full fan beam reconstruction algorithm which is employed to process the weighted partial fan beam data is very sensitive to systematic errors which cannot be corrected by calibration of the detectors. This reconstruction algorithm is most sensitive to errors along the fan axis (axis 75 in Figure AH) so that errors in the data received by the detector along such axis will most strongly affect the appearance of the reconstructed image of the object section. Accordingly, we have recognized that weighting partial fan beam data using a function W(x) of the form shown in FIG. AI results in a moderate improvement in the quality of the reconstructed image (relative to the quality that can be obtained using the function W(x) shown in Figure AH).

After the partial fan data is weighted using function W(x) in the described manner, an image of the scanned object section is reconstructed therefrom in the same manner as from full fan data in accordance with the correlation and back projection algorithms disclosed in referenced U.S. Patent 4,149,249. Thus, the invention permits reconstruction of an image from measured partial fan beam data without reordering the data to simulate data from parallel radiation rays (such a reordering operation requires an additional interpolation step, and correspondingly introduces undesirable signal processing artifacts into the final image).

Consistent with the teaching of U.S. 4,149,249, we prefer that computer 50a and processor 50b are programmed to implement the back projection algorithm in the following manner. For each projection, a

conventional convolution and back projection operation (of normal complexity, and hence of an ordinary number of processing steps) is performed on the detector data to generate a principle set of data. The principal set of data represents the back projection of the detector data onto a "principle line" through the isocenter. The back projections of the data onto the remaining image pixels (specifically, onto lines of pixels parallel to the principle line) are then performed in a streamlined manner involving multiplication of the principle set of data by a multiplicative constant (a different multiplicative constant is used for each line of pixels parallel to the principal line). This implementation is preferred since, in an array processor, it is much more efficient to implement multiplication by a parameter than to implement division by a parameter (it will be appreciated that division is required to generate the principle set of data).

In order to vary the size of the fan (the ratio M/L), IIT 40 should be translated tangentially relative to beam source 30 and object 32 (i.e., along a tangent to the circular gantry rotation path), for example by detector assembly positioning and translation unit 41 shown in Figure M. Thus, to decrease the fan size, IIT face 24 should be translated in the direction of arrow T in Figure E. To increase the fan size, IIT face 24 should be translated in the direction opposite arrow T in Figure E.

This invention is not limited to the preferred embodiment and alternatives heretofore described, to which variations and improvements may be made, without departing form the scope of protection of the

present patent and true spirit of the invention, the characteristics of which are summarized in the following claims.

What is claimed is:

1. A computed tomography method, including the steps of:

(a) sequentially projecting partial fan beams through an object to a detector means, wherein each of the partial fan beams has a different orientation axis;

(b) measuring a set of detector data for each of the projected partial fan beams; and

(c) processing the detector data to generate an image of the object, without reordering the detector data to generate simulated parallel beam data representing detector response to projections of parallel ray beams through the object.

2. The method of claim 1, wherein step (c) includes the steps of:

weighting each said set of detector data to simulate full fan beam data; and

processing the weighted detector data in a processor programmed to reconstruct full fan beam data to generate the image of the object.

3. The method of claim 2, wherein each said set of detector data includes data values representing detectors, wherein each of the partial fan beams has a width M, and wherein step (c) includes the steps of:

for each said set of detector data, multiplying the data values by weighting factors determined by a weighting function W(x), wherein the weighting function W(x) is defined over a range from x = 0 through x = L, where L is the width of a full fan beam spanning the object, and wherein the weighting

function (W(x) satisfies the conditions W(0) = 0; W(L) = 2; and W(x) = 2 - W(L -x).

4. The method of claim 3, wherein the weighting function W(x) is smooth and slowly varying.

5. The method of claim 3, wherein the weighting function W(x) is a continuous function having a first portion equal to zero in the range from x = 0 to x = L-M, a sinusoidal portion in the range from x = L-M to x = M, and a third portion equal to two in the range from x = M to x = L.

6. The method of claim 3, wherein the weighting function W(x) has a first end portion with constant value zero, a second end portion with constant value two, and a middle portion which increases in value from zero to two over its range.

7. The method of claim 6, wherein the range of the middle portion increases with increasing value of the ratio M/L.

8. A computed tomography apparatus, including:

a means for sequentially projecting partial fan beams through an object, each of the partial fan beams having a different orientation axis;

a detector means for receiving the projected partial fan beams and generating a set of detector data for each of the projected partial fan beams; and

a means for processing the detector data to generate an image signal representing the object, without reordering the detector data to generate simulated parallel beam data representing detector

response to projections of parallel ray beams through the object.

9. The apparatus of claim 8, also including a means for displaying the image signal.

10. The apparatus of claim 8, wherein the processing means includes:

a means for weighting each said set of detector data so as to simulate full fan beam data; and

a means for processing the weighted detector data to generate the image signal, wherein the processing means is programmed to perform a full fan beam data reconstruction process on the weighted detector data.

11. The apparatus of claim 10, wherein each said set of detector data includes data values representing detectors, wherein each of the partial fan beams has a width M, and wherein the weighting means includes:

means for multiplying the data values for each set of detector data by weighting factors determined by a weighting function W(x), wherein the weighting function (W(x) is defined over a range from x = 0 through x = L, where L is the width of a full fan beam spanning the object, and wherein the weighting function (W(x) satisfies the conditions W(0) = 0; W(L) = 2; and W(x) = 2 - W(L -x).

12. The apparatus of claim 11, wherein the weighting function W(x) is smooth and slowly varying.

13. The apparatus of claim 11, wherein the weighting function W(x) is a continuous function having a first portion equal to zero in the range from x = 0 to x =

L-M, a sinusoidal portion in the range from x = L-M to x = M, and a third portion equal to two in the range from x = M to x = L.

14. The apparatus of claim 11, wherein the weighting function W(x) has a first end portion with constant value zero, a second end portion with constant value two, and a middle portion which increases in value from zero to two over its range.

15. The apparatus of claim 14, wherein the range of the middle portion increases with increasing value of the ratio M/L.

16. A computed tomography apparatus for imaging an object, including:

a source of fan beam radiation;

a detector array positioned to receive partial fan beam radiation that has propagated through a portion of the object;

means for rotating the source and the detector array along a circular path about an isocenter of the object;

a processing means for receiving partial fan beam data from the detector array, wherein the processing means is programmed to process the partial fan beam data to generate an image signal representing the object, without reordering the partial fan beam data to generate simulated parallel beam data representing detector response to projections of parallel ray beams through the object.

17. The apparatus of claim 16, wherein the detector array includes a main array and an extension array.

18. The apparatus of claim 16, wherein the partial fan beam radiation that has propagated through the portion of the object has a fan beam width, and also including:

means for translating the detector array tangentially with respect to the circular path and relative to the object, in order to vary the fan beam width.

19. The apparatus of claim 16, also including:

means for positioning the detector array to receive full fan beam radiation that has propagated through the object; and

wherein the processing means is programmed to operate in a both full fan mode and a partial fan mode,

wherein, in the full fan mode, the processing means receives full fan beam data from the detector array and generates therefrom a full fan beam image signal representing the object, and

wherein, in the partial fan mode, the processing means receives and processes the partial fan beam data from the detector array to generate a partial fan beam image signal representing the object, without reordering the partial fan beam data to generate simulated parallel beam data representing detector response to projections of parallel ray beams through the object.

20. The apparatus of claim 19, wherein, in the partial fan beam processes the partial fan beam data in accordance with a full fan beam reconstruction operation, and the processing means weights the partial fan beam data to simulate full fan beam data.

21. The apparatus of claim 19, also including:

means for positioning the detector array symmetrically with respect to the isocenter when the processing means is in the full fan mode, and for positioning the detector array asymmetrically with respect to the isocenter when the processing means is in the partial fan mode.

22. A computed tomography method, including the steps of:

(a) sequentially projecting partial fan beams through an object to a detector means, wherein each of the partial fan beams has a different orientation axis;

(b) measuring a set of detector data for each of the projected partial fan beams, wherein each said set of detector data includes data values representing detectors;

(c) weighting each said set of detector data to simulate full fan beam data; and

(d) supplying the weighted detector data to a processor programmed to reconstruct full fan beam data, and processing the weighted detector data in the processor in the same manner as if it were full fan beam data, to generate an image of the object.

23. The method of claim 22, wherein each of the partial fan beams has a width M, and wherein step (c) includes the steps of:

for each said set of detector array data, multiplying the data values by weighting factors determined by a weighting function W(x), wherein the weighting function (W(x) is defined over a range from x = 0 through x = L, where L is the width of a full fan beam spanning the object, and wherein the

weighting function ($W(x)$ satisfies the conditions $W(0) = 0$; $W(L) = 2$; and $W(x) = 2 - W(L - x)$.

24. The method of claim 23, wherein the weighting function $W(x)$ is smooth and slowly varying.

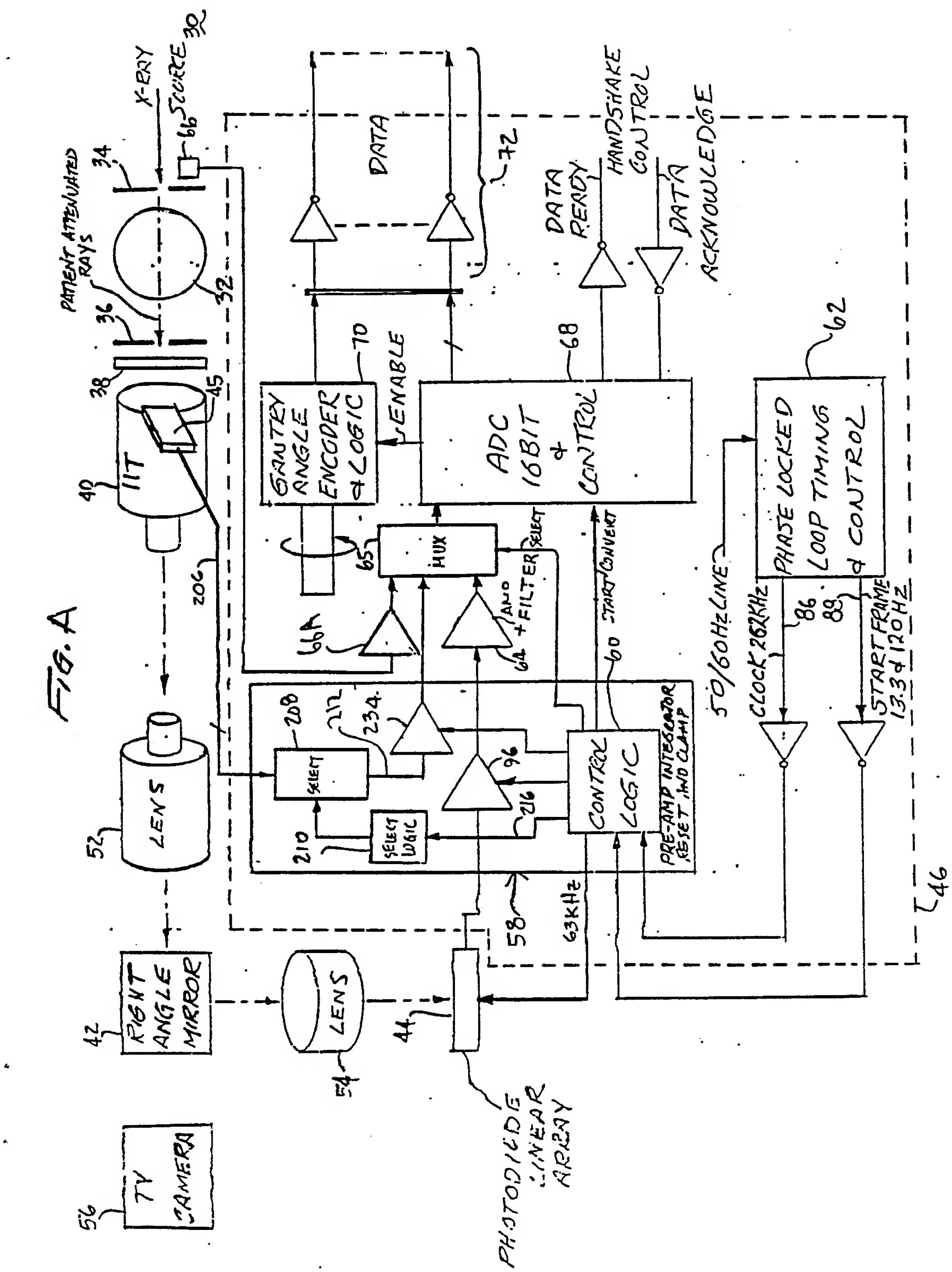

X-RAY SOURCE
30
66
34
PATIENT ATTENUATED RAYS
36
32
38
45
40
IIT
DATA
72
DATA READY
HANDSHAKE CONTROL
DATA ACKNOWLEDGE
70
GANTRY ANGLE ENCODER & LOGIC
ENABLE
68
ADC 16BIT & CONTROL
62
PHASE LOCKED LOOP TIMING & CONTROL
MUX
SELECT
65
AMP + FILTER
START CONVERT
66A
64
60
50/60Hz LINE
CLOCK 262KHz
86
88
START FRAME 13.3 & 120HZ
206
208
212
234
SELECT
96
216
CONTROL LOGIC
SELECT LOGIC
210
PRE-AMP INTEGRATOR RESET AND CLAMP
46
52
LENS
58
63KHz
42
RIGHT ANGLE MIRROR
LENS
54
44
PHOTODIODE LINEAR ARRAY
56
TV CAMERA

FIG. A

FIG. 9

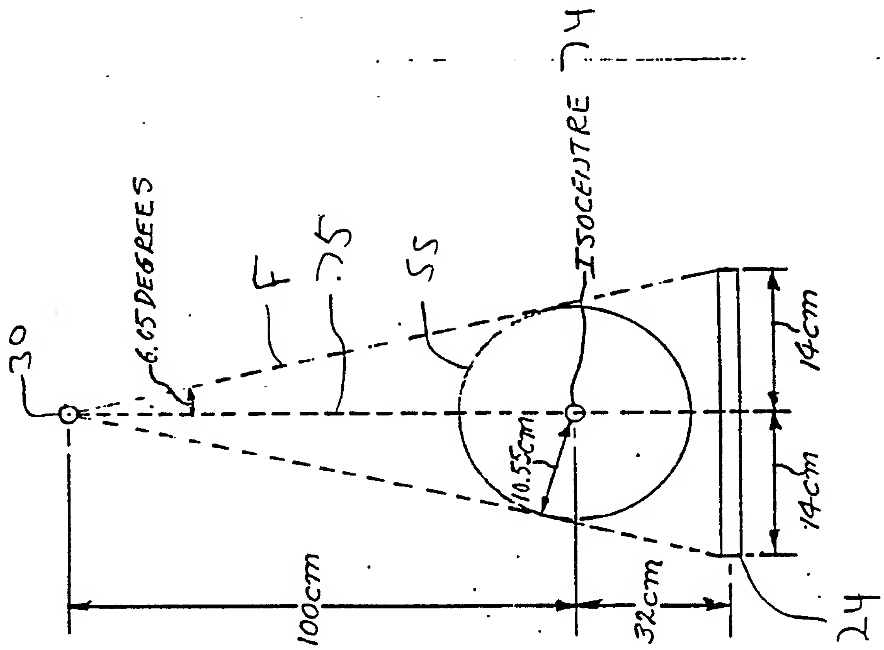

30
6.65 DEGREES
F
75
55
ISOCENTRE 74
10.55cm
14cm
14cm
100cm
32cm
24

FIG. 8

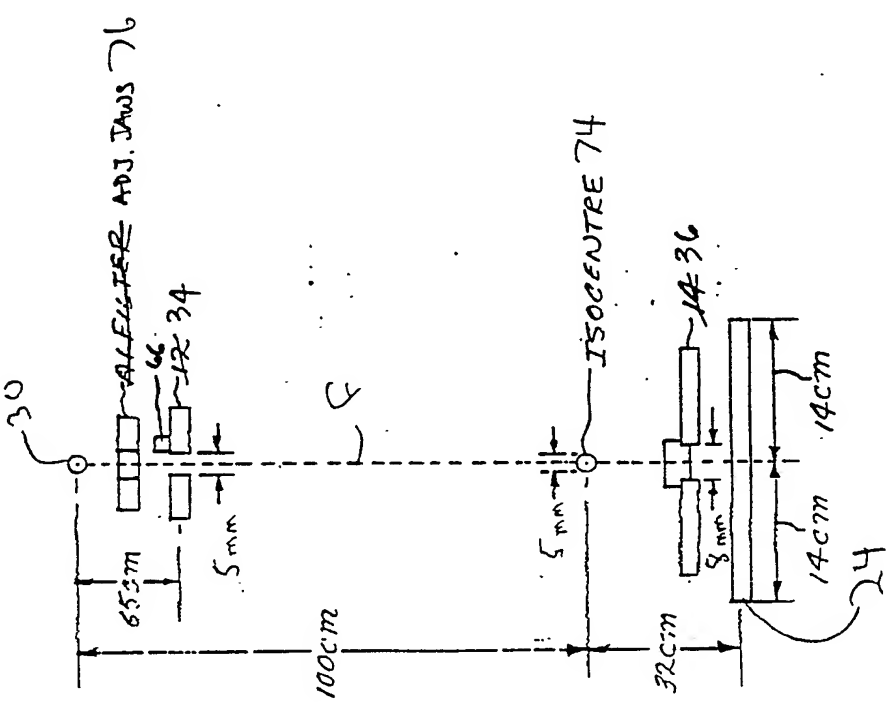

30
ADJ. JAWS 76
34
36
F
ISOCENTRE 74
5mm
5mm
8mm
14cm
14cm
100cm
32cm
24

FIG. E

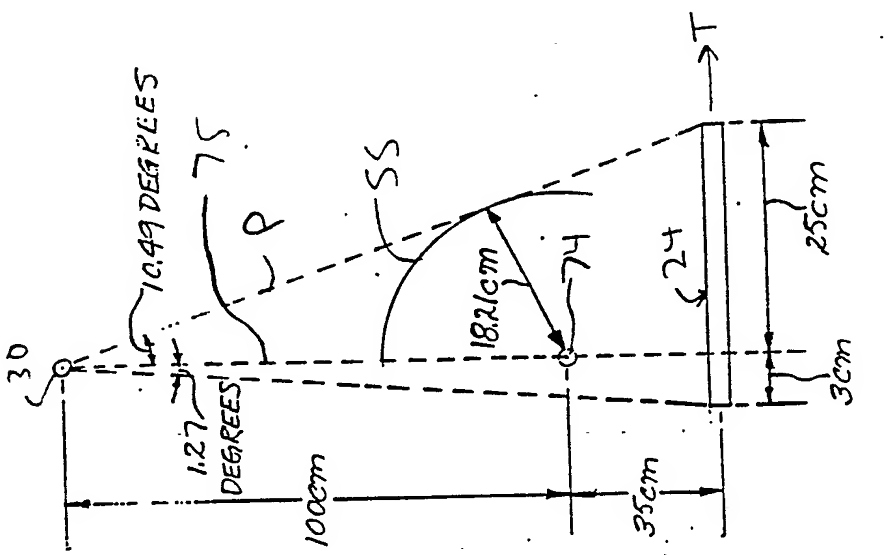

10.49 DEGREES
75
55
P
30
18.21cm
74
24
T
25cm
3cm
1.27 DEGREES
100cm
35cm

FIG. D

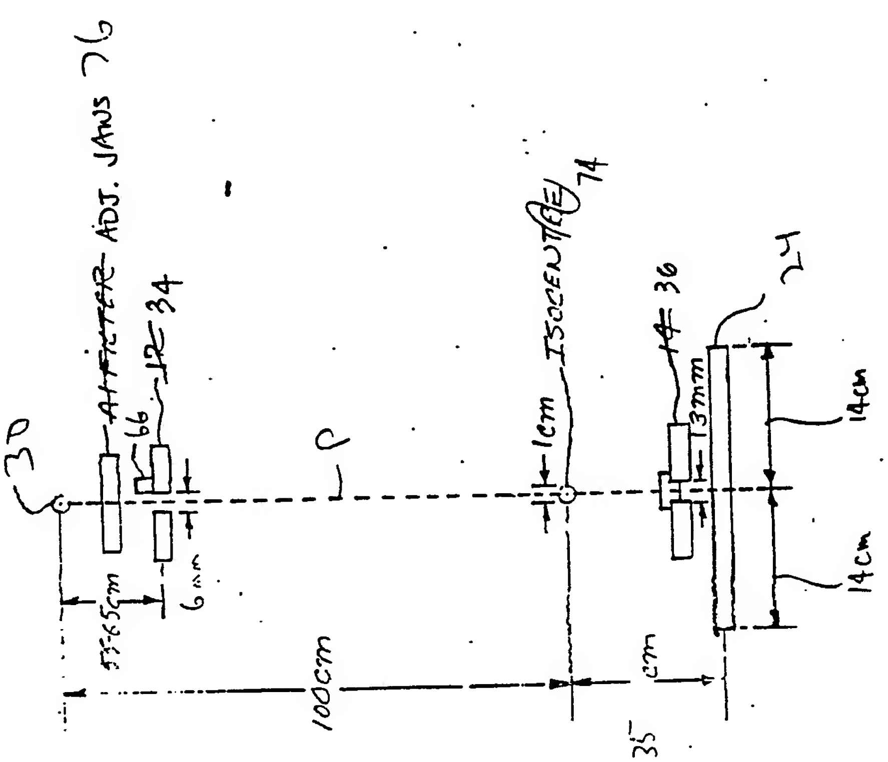

ADJ. JAWS 76
ISOCENTRE 74
36
34
66
P
30
1cm
3mm
24
14cm
14cm
53-65cm
6mm
100cm
35 cm

FIG. 8

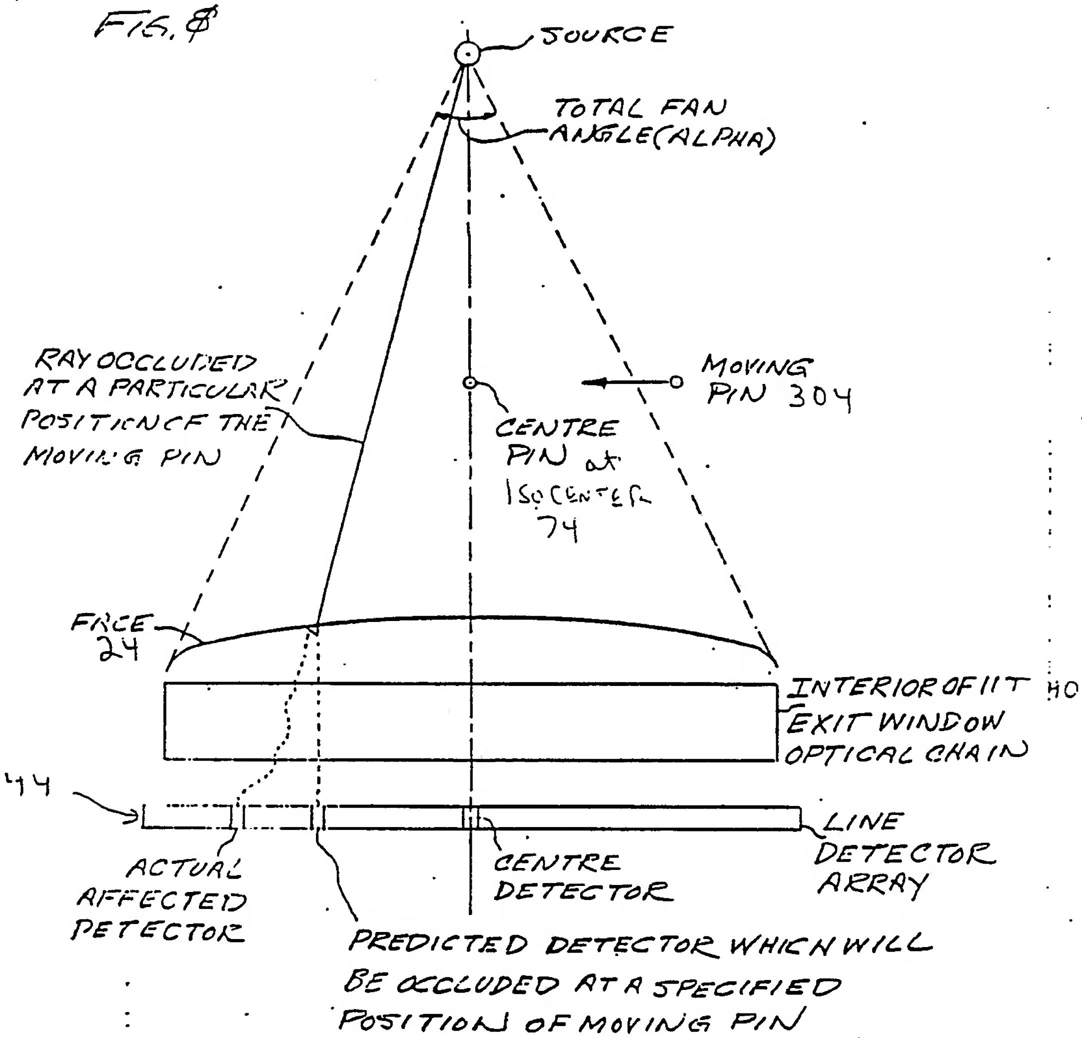

SOURCE
TOTAL FAN ANGLE(ALPHA)
RAY OCCLUDED AT A PARTICULAR POSITION OF THE MOVING PIN
MOVING PIN 304
CENTRE PIN at ISOCENTER 74
FACE 24
INTERIOR OF IIT 40
EXIT WINDOW
OPTICAL CHAIN
44
LINE DETECTOR ARRAY
ACTUAL AFFECTED DETECTOR
CENTRE DETECTOR
PREDICTED DETECTOR WHICH WILL BE OCCLUDED AT A SPECIFIED POSITION OF MOVING PIN

FIG. I

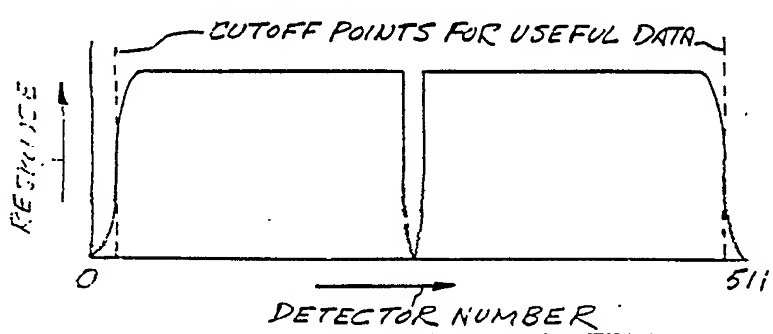
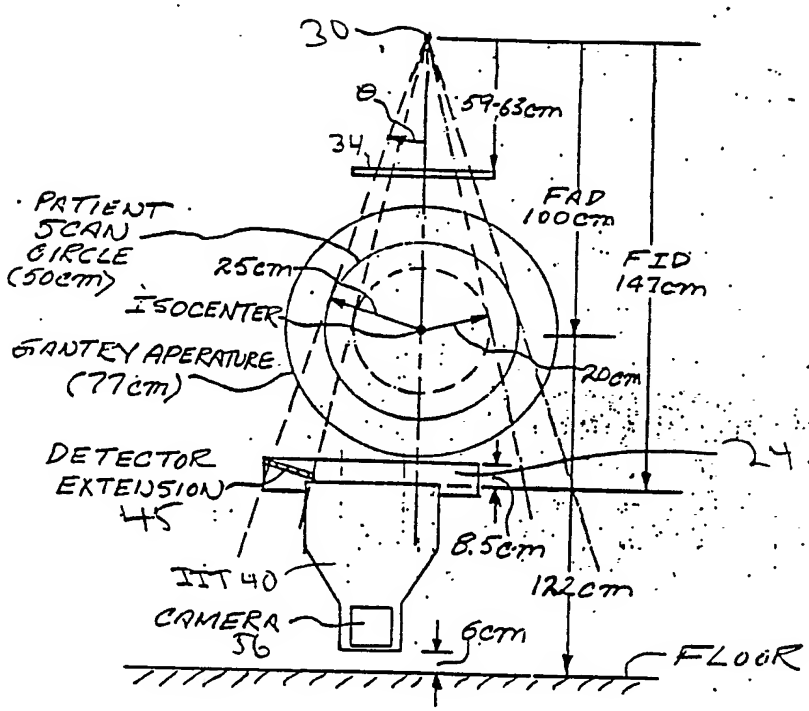

CUTOFF POINTS FOR USEFUL DATA
RESPONSE
0
511
DETECTOR NUMBER

FIG. 1L


30
θ
59-63cm
34
FAD
100cm
PATIENT
SCAN
CIRCLE
(50cm)
25cm
FID
147cm
ISOCENTER
GANTRY APERATURE
(77cm)
20cm
DETECTOR
EXTENSION
45
24
8.5cm
IIT 40
122cm
CAMERA
56
6cm
FLOOR

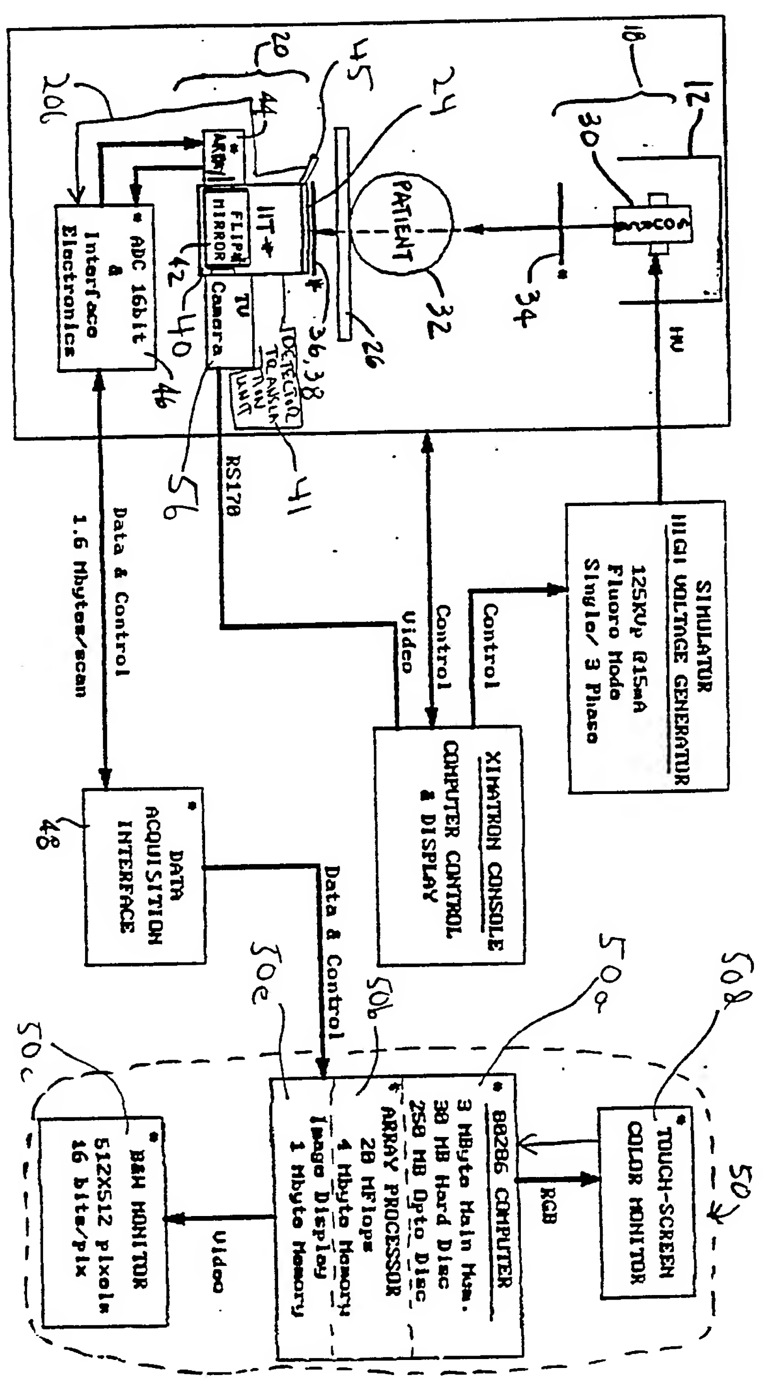

SIMULATOR
HIGH VOLTAGE GENERATOR
125KVp @15mA
Fluoro Mode
Single/ 3 Phase
HV
PATIENT
IIT
FLIP MIRROR
TV Camera
ADC 16bit
&
Interface
Electronics
DETECTOR TRANSLATION UNIT
Control
Control
Video
RS170
XIMATRON CONSOLE
COMPUTER CONTROL
& DISPLAY
Data & Control
1.6 Mbytes/scan
DATA
ACQUISITION
INTERFACE
Data & Control
80286 COMPUTER
3 MByte Main Mem.
30 MB Hard Disc
250 MB Opto Disc
ARRAY PROCESSOR
20 MFlops
4 MByte Memory
Image Display
1 Mbyte Memory
RGB
TOUCH-SCREEN
COLOR MONITOR
Video
B&W MONITOR
512X512 pixels
16 bits/pix
12
18
30
34
32
24
45
20
44
26
36,38
40
42
46
41
56
206
48
50
50a
50b
50c
50d
50e

FIG.M

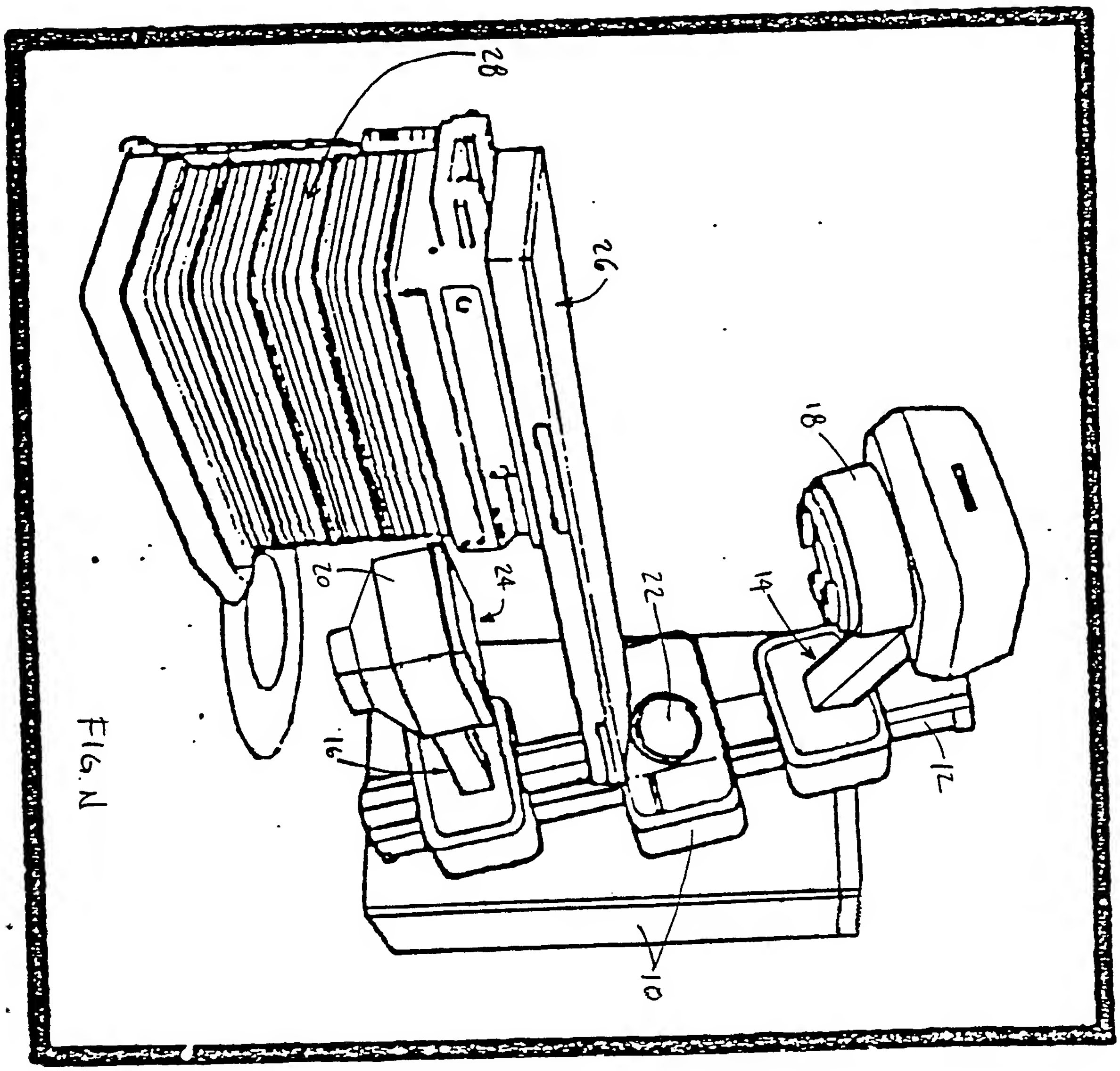


FIG. N

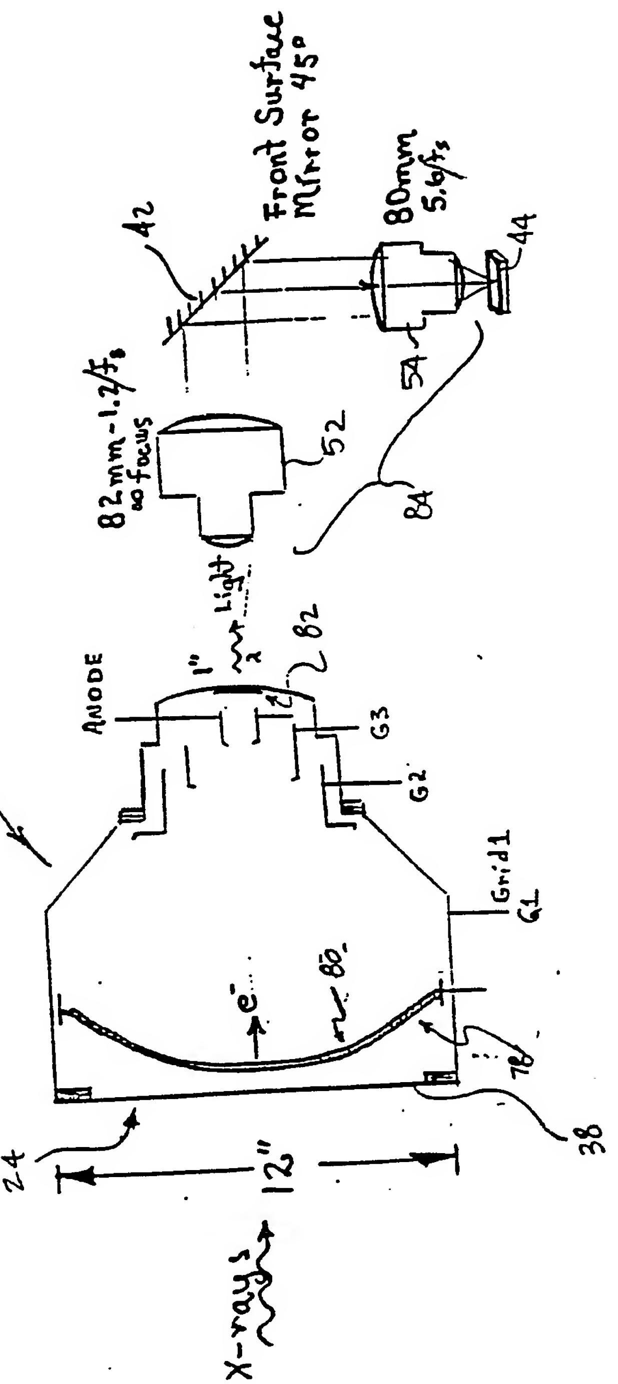

Front Surface
Mirror 45°
42
80mm
5.6/fs
44
82mm-1.2/f8
∞ focus
52
54
84
Light
82
1"
ANODE
40
G3
G2
Grid1
G1
e-
80
78
38
24
12"
X-rays

FIG. 0

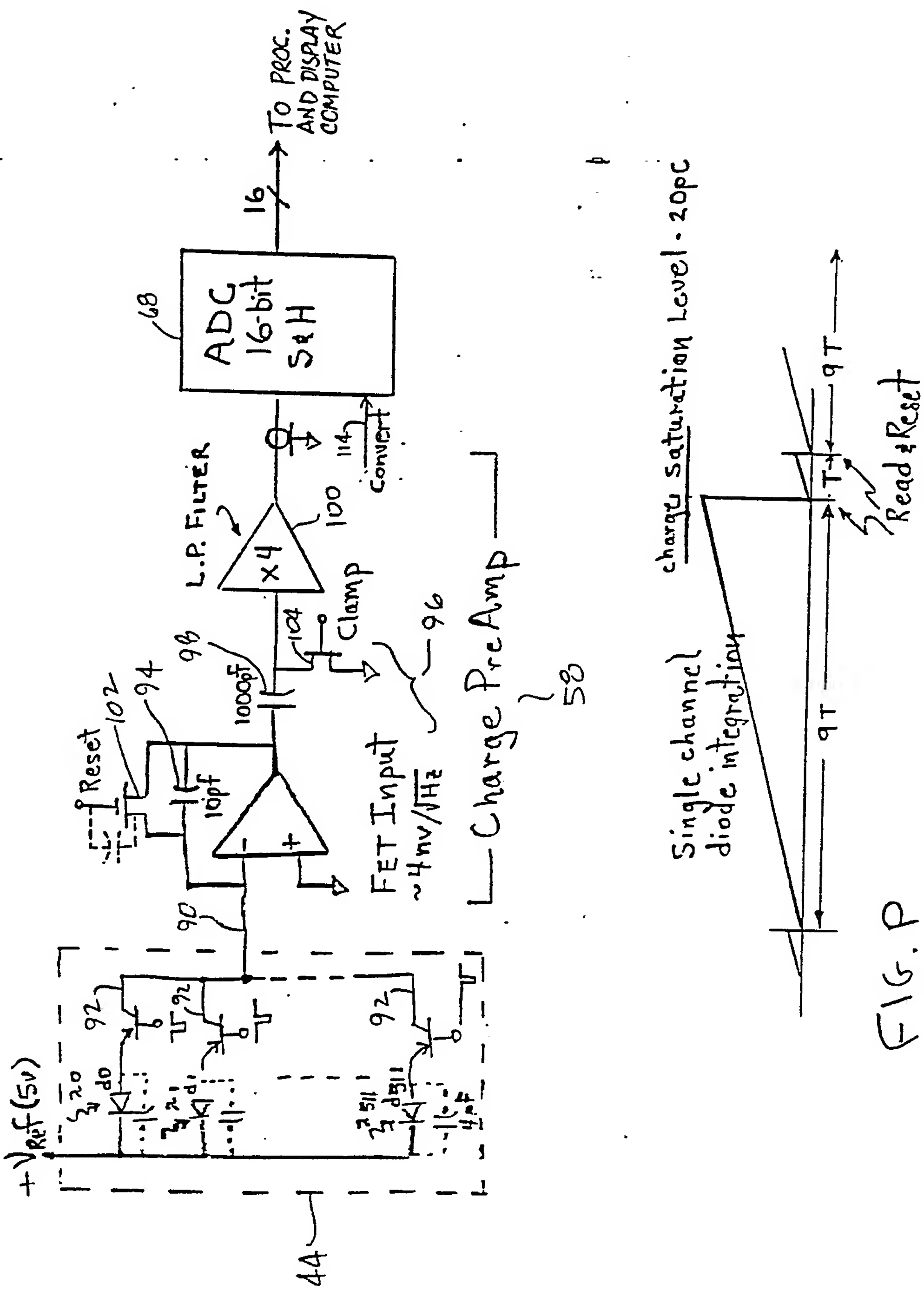

+V Ref (5v)
44
92
d0
d1
d311
4pF
90
Reset
102
94
10pf
98
1000pf
104
Clamp
96
FET Input
~4nv/√Hz
Charge Pre Amp
58
L.P. Filter
x4
100
114
Convert
68
ADC
16-bit
S&H
16
TO PROC. AND DISPLAY COMPUTER
Single channel diode integration
charge saturation Level - 20pc
9T
T
Read & Reset

FIG. P

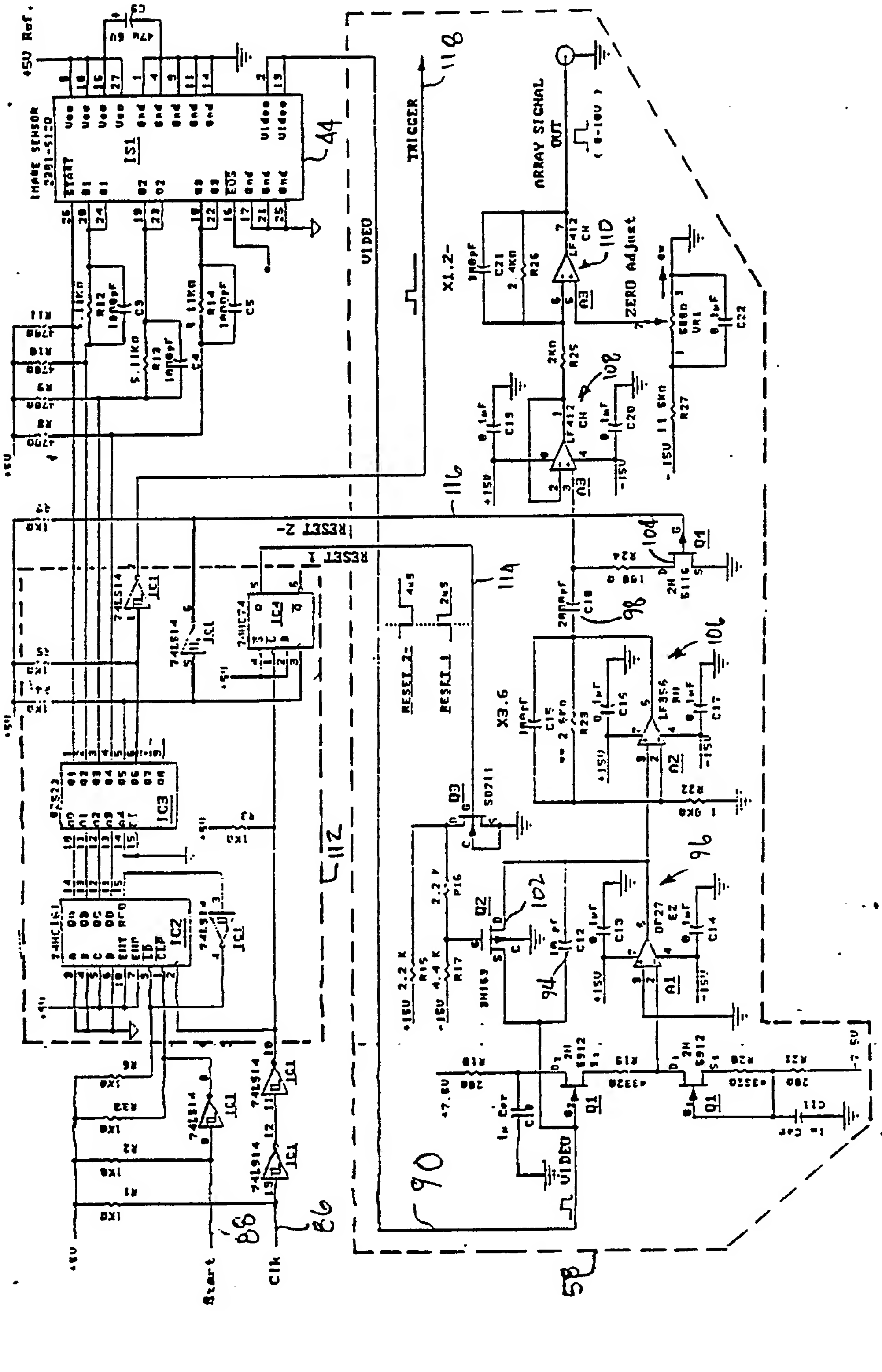


FIG. Q

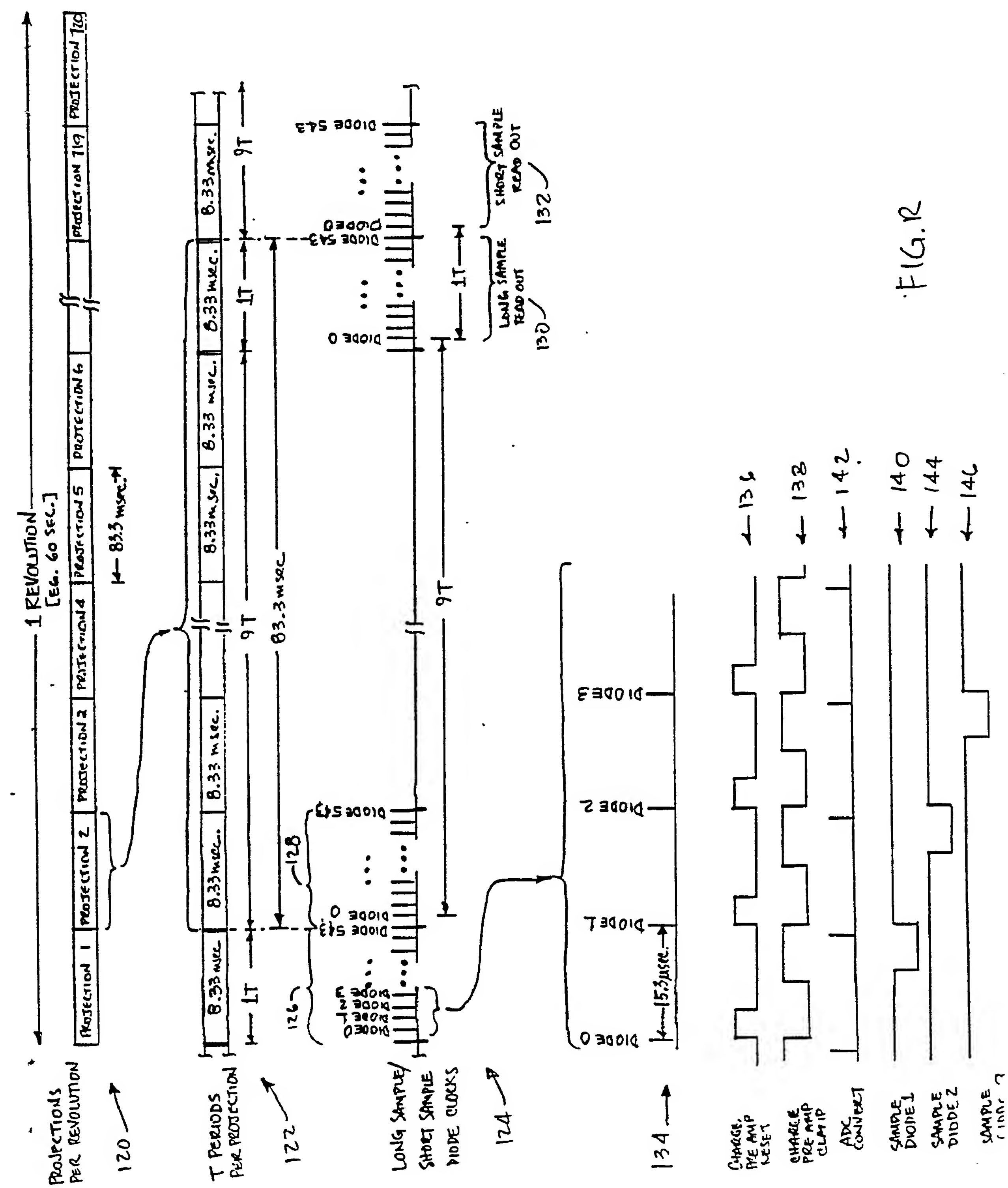

1 REVOLUTION
[EG. 60 SEC.]
PROJECTIONS PER REVOLUTION
120
T PERIODS PER PROJECTION
122
LONG SAMPLE/ SHORT SAMPLE DIODE CLOCKS
124
134
LONG SAMPLE READ OUT
130
SHORT SAMPLE READ OUT
132
CHARGE PRE AMP RESET
CHARGE PRE AMP CLAMP
ADC CONVERT
SAMPLE DIODE 1
SAMPLE DIODE 2

FIG. 12

FIG. S

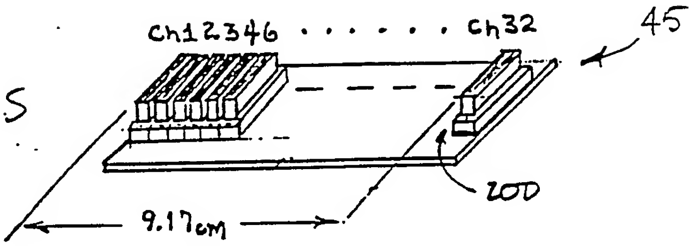


FIG. T

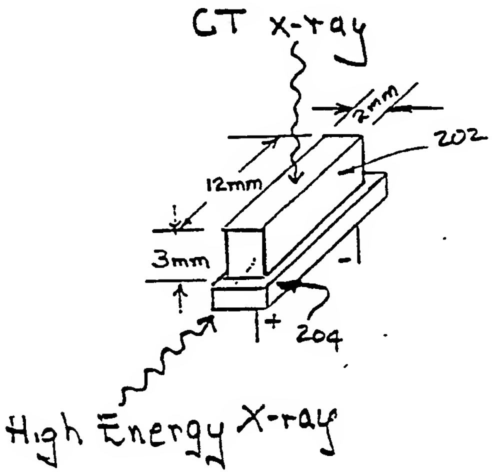
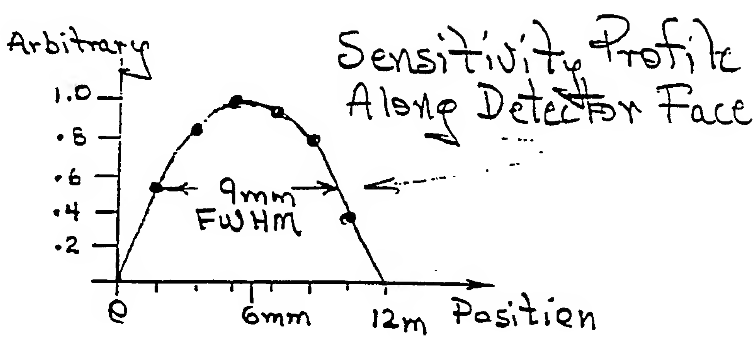


FIG. U



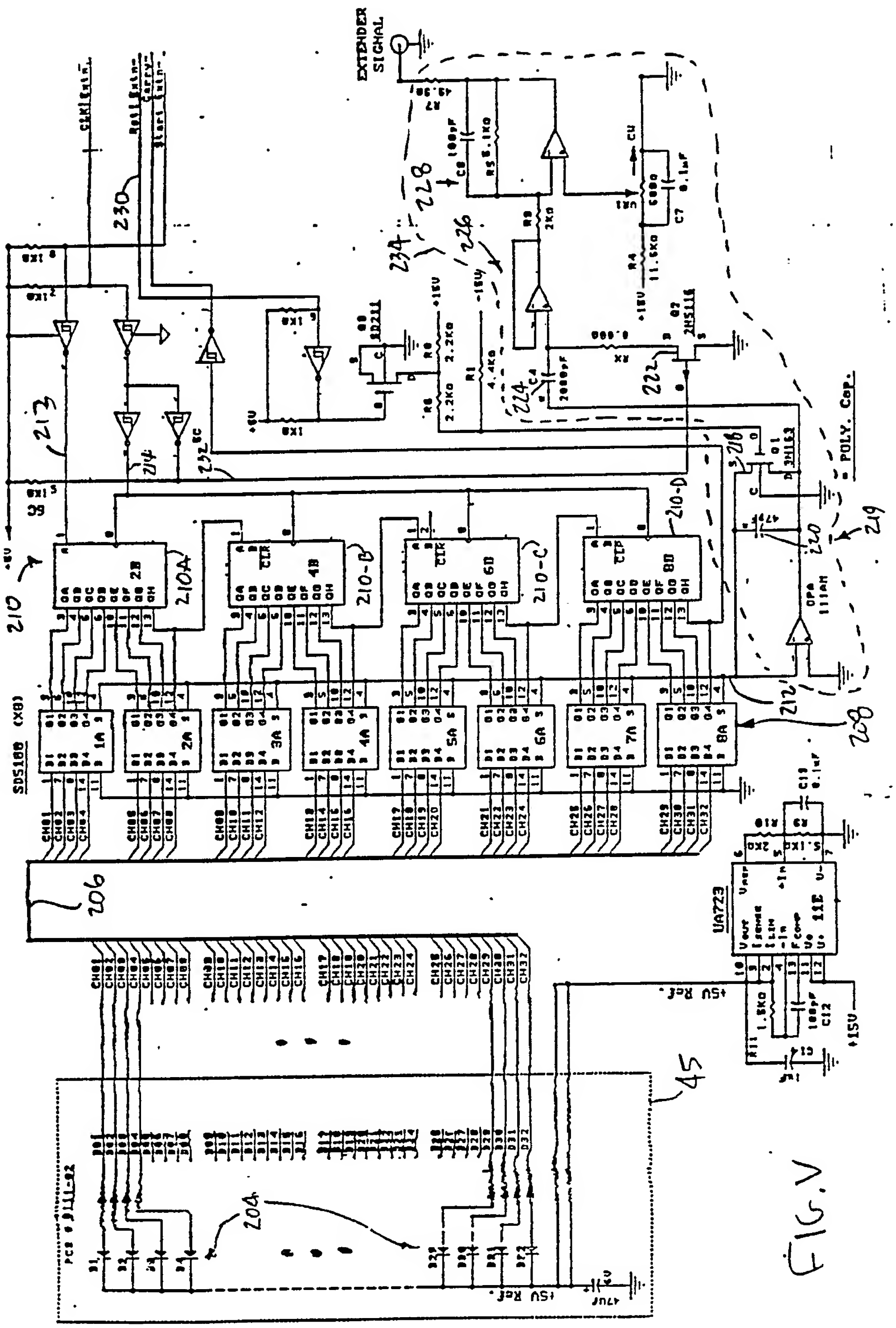


FIG. V

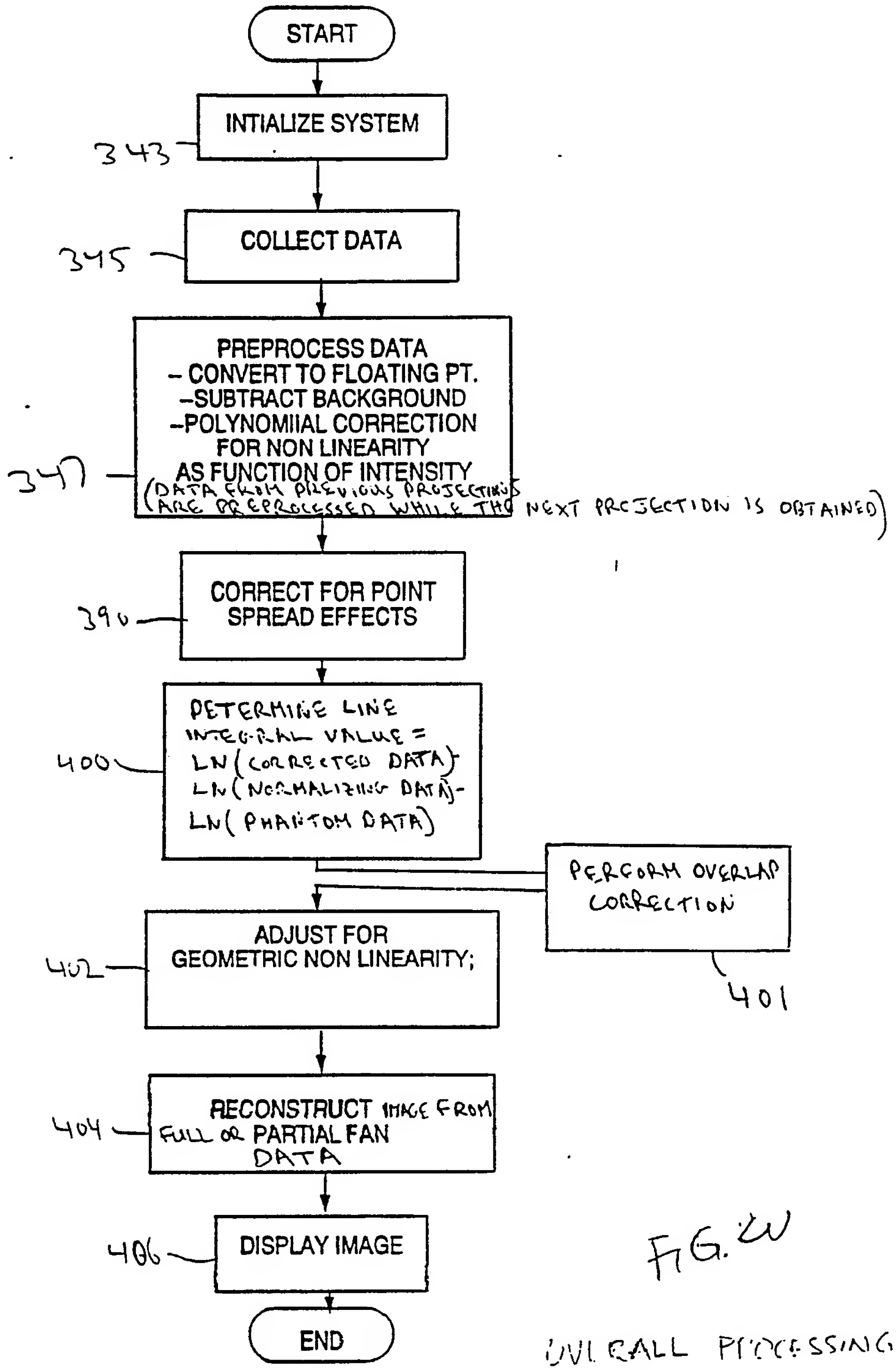


FIG. 20

OVERALL PROCESSING FLOW

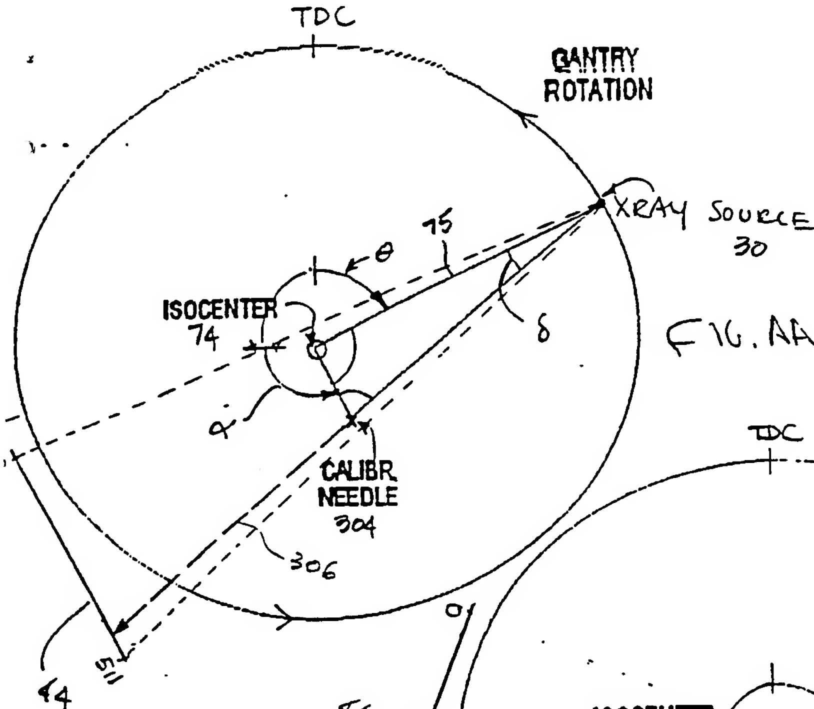


FIG. AA

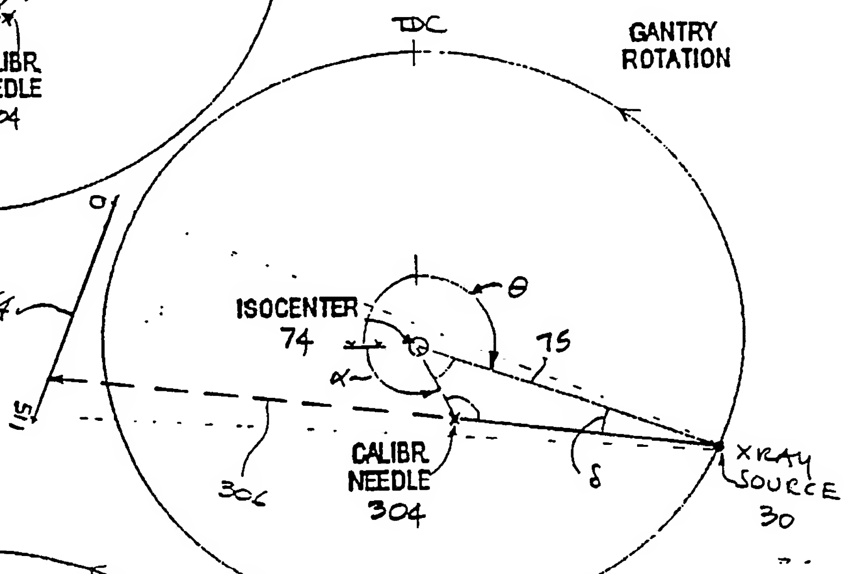


FIG. AB

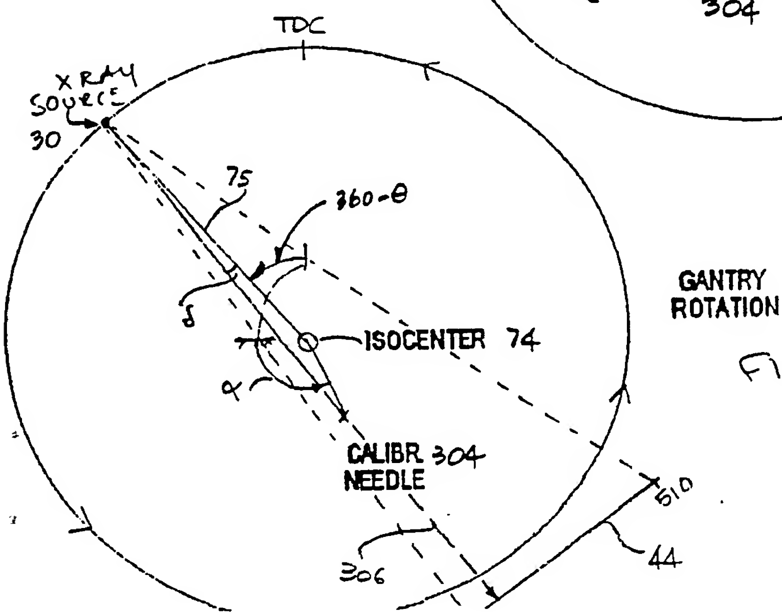


FIG. AC

DETECTOR NUMBER	ACTUAL ANGLE FOR RESPONSE	DESIRED ANGLE FOR RESPONSE ($\Delta = 0.048°$)
0	-11.995	-12°
1	-11.899	-11.952
2	-11.820	-11.904
3	-11.728	-11.856
⋮		⋮
510	+11.870	+11.952
511	+11.970	+12°

FIG. AD.

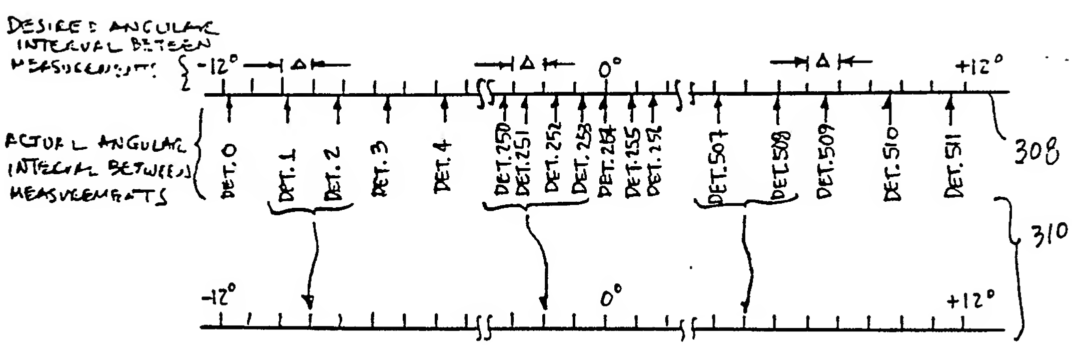
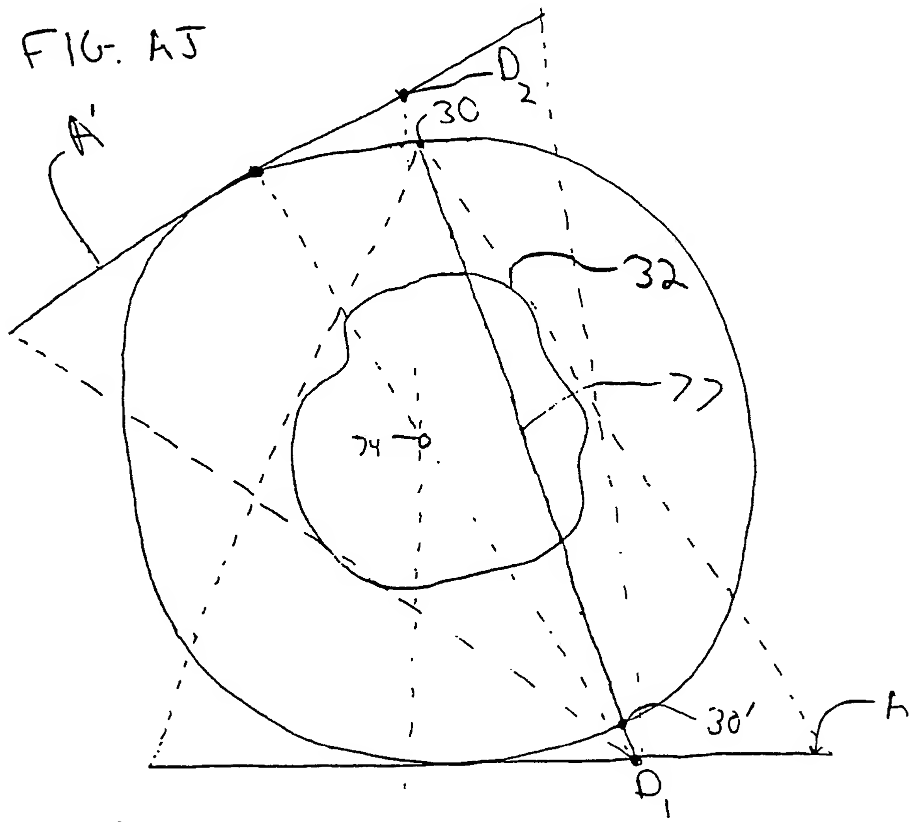
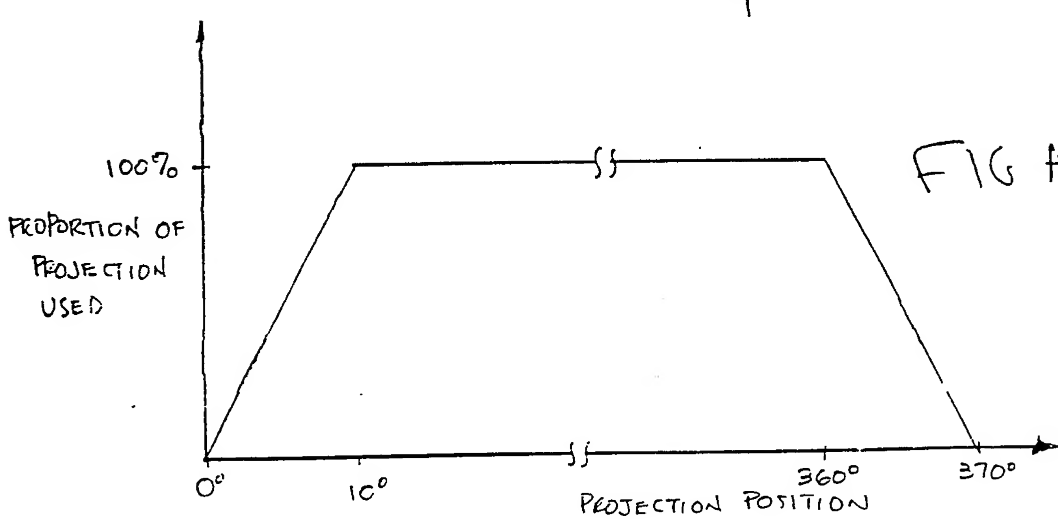


FIG. AE

FIG. AJ



A'
D2
30
32
77
74
30'
A
D1



100%
PROPORTION OF PROJECTION USED
0°
10°
360°
370°
PROJECTION POSITION

FIG AG

FIG. AH

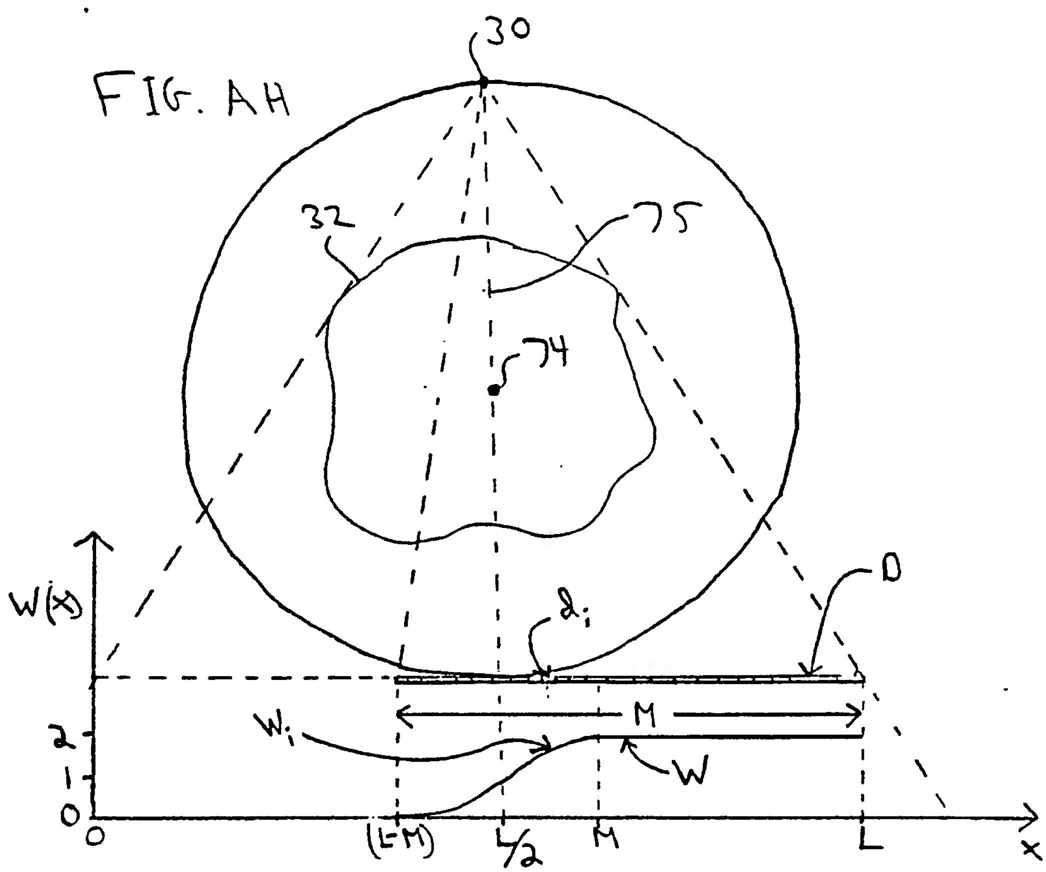
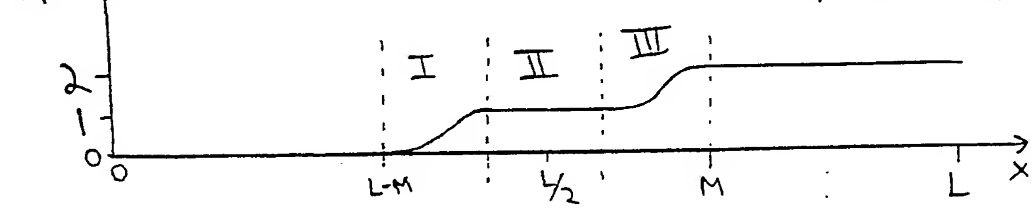


FIG. AI



INTERNATIONAL SEARCH REPORT

International Application No. PCT/US91/04776

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all) [6]

According to International Patent Classification (IPC) or to both National Classification and IPC

IPC(5): G06F 15/42
U.S. CL: 364/413.17

II. FIELDS SEARCHED

Minimum Documentation Searched [7]

Classification System	Classification Symbols
U.S. CL.	364/413.14, 413.16-413.22; 378/11-14,99

Documentation Searched other than Minimum Documentation to the Extent that such Documents are included in the Fields Searched [8]

III. DOCUMENTS CONSIDERED TO BE RELEVANT [9]

Category *	Citation of Document, [11] with indication, where appropriate, of the relevant passages [12]	Relevant to Claim No. [13]
Y	US,A 4,149,079 (BEN-ZEEV ET AL.), 10 April 1979, see figures and column 2, lines 38-40.	1,8,16, 22
Y	US,A 4,066,900 (LEMAY), 3 January 1978, see figures and column 4, lines 4-6.	1,8,16, 22
Y	US,A 4,091,285 (LOGAN ET AL.), 23 May 1978, see figures and column 5, lines 1-3.	1,8,16, 22
Y	US,A 4,149,248 (PAVKOVICH), 10 April 1979, see column 8, lines 3-5.	1,8,16, 22
Y	US,A 4,149,247 (PAVKOVICH), 10 April 1979, see column 8, lines 9-11.	1,8,16, 22
A	US,A 4,191,892 (HUANG ET AL.) 4 March 1980, see figure 8.	2-7, 10-15, 23-24

* Special categories of cited documents: [10]

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier document but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention

"X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step

"Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

IV. CERTIFICATION

Date of the Actual Completion of the International Search	Date of Mailing of this International Search Report
21 October 1991	05 DEC 1991
International Searching Authority	**Signature of Authorized Officer** Ngoc-Ho Nguyen
ISA/US	NGUYEN NGOC-HO for DAVID HUNTLEY INTERNATIONAL DIVISION

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